Chapter 5: Cyborg Technology - Biomimetic Orthotic and Prosthetic Technology

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5.0 Introduction

A long standing goal in engineering is to exploit the unique designs of the body to guide the development of anthropomorphic artificial appendages that exhibit human-like stability, strength and speed in a variety of natural environments. Although tremendous technological progress has been made since the days of the wooden peg leg, contemporary orthotic and prosthetic (O&P) limbs cannot yet perform as well as their biological counterparts, whether in terms of stability, fatigue-life or speed (Popovic & Sinkjaer 2000). However, in the next several decades continued advances in human-machine neural interfaces, muscle-like actuators and biomimetic humanoid control schemes may result in dramatic improvements in the quality of life of the physically challenged. In this chapter, we review key research areas relevant to the O&P field. By way of case study, we describe both artificial and actin-myosin based muscle actuators, control methodologies that exploit principles of biological movement, and device architectures that resemble the body's own skeletal design. We limit our discussions to external devices, specifically designed for human rehabilitation, that attach to human arms and legs. Described are orthoses that attach *in parallel* to human limbs for the treatment of limb dysfunction, and prostheses that attach *in series* to limbs for the treatment of limb amputation. After completing the chapter, you will know the brief history of O&P appendages, from the crude peg leg used by the Romans to contemporary microprocessor-controlled artificial limbs. You will also learn of research areas that are actively being studied that may prove critical to the next generation of O&P technology.

5.1 A Brief History of O&P Technological Development

We know from Egyptian stelae (2500 BCE) and from early Roman mosaics that prostheses and simple walking aids (orthoses) have been used during much of recorded history. People in parts of the world still use a head-high wooden stick to vault over on the side of their non-functional limb when they walk, much as some disabled Egyptians did thousands of years ago. Simple wooden canes must be nearly as old as human kind itself. Wooden peg legs have been effective aids to walking for thousands of years. Until the 20th Century, wood and leather were the favorite composite materials in O&P devices. Paintings of Brueghel from the 16th Century show clearly the plight of persons without limbs or with dysfunctional limbs as a result of polio or cerebral palsy. Most of their locomotory aids were fashioned from wood and leather, perhaps by themselves.

Wars and conflicts have inevitably stimulated developments in O&P technology, and the armor makers of the medieval era were early O&P practitioners. The noble German knight, Götz von Berlichingen, remarked in Goethe's play *The Iron Hand*, that his iron hand had served him better in the fight than ever did the original of flesh. Ambroise Paré (1510-1590), a French army surgeon can rightly be called the father of amputation surgery and prosthetics. He developed the ligature, which eliminated searing the residual limb to stop bleeding. He used site selection to try to produce limbs that were as useful as possible, and he designed prostheses and followed the outcome of his patients.

Not all warriors wore prostheses. The next time you are in Trafalgar Square in London, observe the statue of Viscount Horatio Nelson. Nelson lost his arm above the elbow at Tenerife, lost sight in his right eye on Corsica, received a severe head injury at Alexandria, and watched his greatest naval victory at Trafalgar while propped up on deck with a fatal spinal cord injury.

His only rehabilitation aid was the "Nelson knife", now frequently called a rocker knife, which remains even today one of the best eating aids for persons with only one arm.

The Napoleonic wars played their part in prosthetics development, mostly in France and in England. Lord Uxbridge, Wellington's cavalry officer at Waterloo became the wearer an above knee prosthesis that became known as the Anglesea Leg after the island of Anglesea where Uxbridge resided after the war. It was a unique prosthesis that raised the toe as the knee was flexed in order to reduce stumbling. The concept is still used today. The Anglesea prosthesis, after some changes, was used widely in America by veterans of the Civil War. The enormous number of amputations resulting from the American Civil War established the prosthetics industry in the United States during the late 1800s. However, it was WWI that set the stage for modern prosthetics. Many of the early advances occurred in Germany.

In Zürich, about 1915, a well-known German surgeon, Ferdinand Sauerbruch, worked with Aurel Stodola, a famous turbine engineer and professor of mechanics at the Polytechnic Institute of Zürich to produce a hand prosthesis that was controlled and powered directly from surgically prepared muscles of the residual limb. The surgical technique developed to achieve this biological control mechanism was called muscle tunnel cineplasty (For more details, read the **Cineplasty** section of this chapter). Sauerbruch was one of the first surgeon/physicians to recommend multidisciplinary scientific and engineering endeavors in the prosthetics/rehabilitation field. After successfully developing the Sauerbruch hand he said, "Henceforth, surgeon, physiologist, and technologist will have to work together." (page 452, The Literary Digest for August 26, 1916). After WWI, American surgeons studied surgical and prosthetic rehabilitation methods in Europe, such as Sauerbruch's tunnel cineplasty and Krukenberg's surgical fashioning of the radius and ulna of the limb of long below elbow amputees into two large "fingers" that could be used effectively for gripping large objects. However, no research and development work was fostered in America after WWI.

World War II mobilized research and development of prostheses all over the world. In America this burst of research activity was stimulated partially by veteran amputees who were languishing in hospitals and who were disappointed by the state of limb prosthetics in 1945. As a consequence of their lobbying, the surgeon general of the Army asked the National Research Council to call a meeting to select which prostheses would be best for the veterans of WWII. This meeting, held in Chicago during January of 1945, produced recommendations for scientific and engineering studies of limb prostheses. From this meeting the first federal grants were issued to promote the science and technology of prostheses and amputation. Early investigations included tours of O&P facilities in many countries. The early studies and the new research were dramatically successful, and the period from 1945 to 1975 was perhaps the most productive period ever in American orthotics and prosthetics. In fact, this period was productive for O&P technology worldwide. The O&P field is international in scope and this brief history has captured only a few happenings in a handful of countries.

5.2 O&P Systems: State of the Art

In this section, we give a brief overview of the state-of-the-art in O&P technology. For upper and lower extremity systems, we describe original 'archetypal' designs from which existing O&P devices have emerged: passive or cosmetic prosthetic devices, body powered prosthetic arm devices, myoelectric systems, dissipative prosthetic knees, energy-storing prosthetic feet, and passive limb orthoses.

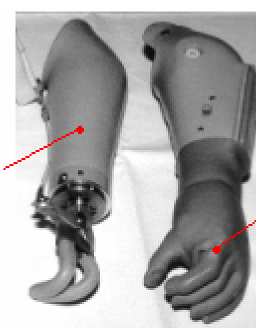
5.2.1 Upper Extremity Prosthetic Systems

Passive or Cosmetic Devices. The passive or cosmetic prosthesis is the oldest archetype, with the first recorded being an artificial hand made for Marcus Sergius in the Punic War (218-201 BC). This device appeared as a gauntleted hand, or armoured glove, and was made by skilled armourers (Banerjee 1982). Although today's passive or cosmetic prostheses differ in that they appear as a static representation of the naked human hand, they are similarly made using craft techniques by highly skilled professionals (Figure 1a). However, although a remarkable static representation of the human hand is achieved, the absence of movement can promote feelings of unease in the observer, as effective 'cosmesis' also includes a naturalness of movement (Reichardt 1978).

Body Powered Devices. The body-powered prosthesis originates from 1812, when Peter Baliff, a Berlin dentist, invented a 'terminal device' that was operated from straps attached to the trunk of the body of the amputee (Banerjee 1982). Current body-powered devices (Figure 1b) are operated using straps that commonly pass over the amputee's shoulders and are operated by biscapula adduction (rounding of the back and shoulders). Fine control is possible as there is a pathway for 'feed-back' or 'prioproception' to an intact joint enabling the amputee to know where the device is 'positional feedback' and how much force is being exerted 'force feedback'. However, the straps required for control are problematic. They make donning and doffing the prosthesis difficult, and may interfere with other garments. Additionally, the movements that are needed for the control of the prosthesis are considered ungainly to some amputees (Whiteley 2000). With this method of control, practically only a single degree of freedom can be controlled, however, catches are often utilized that transfer control from the gripping of the hand to the flexion of the elbow (Bennet Wilson 1989).

1a. A passive prosthesis made at the Northern General Hospital, Sheffield.

1b. A body-powered prosthesis



1c. *A* myoelectric prosthesis

FIGURE 1: Types of prostheses currently available to upper extremity amputees.

Myoelectric Devices. The myoelectric prosthesis or 'myo' was invented in 1948 by Reihold Reiner (Kostuik 1980). This type of prosthesis uses sensors to detect, commonly, a threshold of electromyographic activity to switch an electric motor in the artificial 'hand', and can also be used to switch powered wrist and elbow components. Electromyographic (EMG) activity originates from the depolarisation and repolarisation of the individual muscle cell membranes during muscle activity. Using surface electrodes it is possible to measure these potential differences on surrounding skin (Kostuik 1980). There are many permutations of this control scheme, however, a salient point is that commercial myoelectric prostheses do not operate in a 'volitional' manner. Rather, the amputee is taught to achieve the necessary degrees of muscular contraction corresponding to the threshold levels of electrical activity needed for operation (Scott & Parker 1988). Additionally, the myoelectric control method only provides an 'efferent' signal to the prosthesis from the amputee. There is no 'afferent' signal returned to the amputee to inform what grip strength is being applied or what position the fingers are in; unlike the mechanical connection to the body of the body-powered device. The greatest benefits of myoelectric prostheses are their increased grip strength compared to body-powered devices, and that there is no necessity for the donning of elaborate control straps combined with the fact that the myo often has a more hand-like appearance (Figure 1c) (Datta & Ibbotson 1998).

5.2.2 Lower Extremity Prosthetic Systems

Dissipative Knees. Prosthetic knees in use today typically comprise a hydraulic and/or pneumatic damper that dissipates mechanical energy under joint rotation (Popovic & Sinkjaer 2000). In these devices, fluid is pushed through an orifice when the knee is flexed or extended, resulting in a knee torque that increases with increasing knee angular rate. To control knee damping, the size of the fluid orifice is adjusted. Although for most commercial knees the orifice size is controlled passively when weight is applied to the prosthesis, some contemporary knee systems use a motor to actively modulate orifice size. In Figure 2A, the Otto Bock C-Leg is shown. In this system, hydraulic valves are under microprocessor control using knee position and axial force sensory information (James et al. 1990). Actively controlled knee dampers such as the C-Leg offer considerable advantages over passive knee systems, enabling amputees to walk with greater ease and confidence (Dietl & Bargehr 1997, Kastner et al. 1998).

FIGURE 2: Types of prostheses currently available to lower extremity amputees.

2a. The C-Leg Prosthetic Knee

2b. The Flex-Foot

Energy-Storing Prosthetic Feet. Today's prosthetic ankle-foot systems typically use elastomer bumper springs or carbon composite leaf springs to store and release energy throughout each walking or running step (Popovic & Sinkjaer 2000). In Figure 2B, the Flex-Foot Vertical Shock Pylon System is shown. In this device, carbon composite leaf springs offer considerable heel, toe and vertical compliance to the below-knee prosthesis, enabling leg amputees to move with greater comfort and speed. Although considerable progress has been made in materials and methods, commercially available ankle-foot devices are passive, and consequently, their stiffnesses are fixed and do not change with walking speed or terrain.

5.2.3 Lower and Upper Extremity Orthotic Systems

Limb orthoses (otherwise known as braces, splints, callipers or supports) have historically been made using craft techniques. Orthoses may be used in the postoperative treatment of burns or fractures, whilst other uses include providing support and protection for arthritis joints. A further use of limb orthoses is to add constraint to joints in users who may suffer from neurological disorders (Leonard et al. 1989). While traditionally these devices have been constructed using rigid thermoplastics lined with soft foam, more contemporary designs, especially for the lower limb, are substituting these materials for lightweight composites. Additionally, current fabrication methods are moving from craft intensive techniques to numerically controlled methods (Smith and Burgess 2001). However, despite progress in materials and methods all commercially available orthoses are passive devices. Only recently has interest been renewed in the research into powered orthotic devices (Rahman et al. 2000, Blaya 2002).

5.3 New Horizons for O&P Technology: Merging Body and Machine

Most critical to the advancement of biomimetic O&P mechanisms is the development of actuator technologies that behave like muscle, control methodologies that exploit principles of biological movement, and device architectures that resemble the body's own skeletal design. In this section, we describe several research initiatives that, if successful, may result in more life-like artificial appendages and a greater intimacy between body and O&P device. We discuss upper extremity prostheses that mimic the skeletal articulations of the human forearm, machine actuation using living muscle tissue, O&P control schemes motivated by biomechanical models, and actuated hand prostheses that employ *in vivo* measures of muscle force to modulate prosthesis gripping force.

5.3.1 Biomimetic Structural Design

Critical to the advancement of biomimetic O&P systems is the development of artificial joint architectures that mimic the skeletal articulations of the human body. In this section, we give one example of an O&P mechanism that resembles the body's own structural design. From first principles, the skeletal morphology of the human arm is reconstructed into an external prosthesis. Through the interaction of an artificial radius and ulna, the prosthesis exhibits human-like pronation and supination movements, affording more forearm space for extrinsic hand actuators than is available using bulky wrist rotators typically found in conventional prosthetic systems.

5.3.1.1 Need for a First Principles Approach

From long attendance at an amputee support group, consisting of arm amputees of both genders and of varying ages, the impression gained was of their deep dissatisfaction with currently available prostheses, both in terms of their limited function and often their poor cosmetic appearance (Whiteley 2000). Additionally, it appeared that the poor functionality of existing devices is largely *overcome* by amputees' 'adapting' the methods in which activities of daily living are achieved. Amputees need to carry out normal activities of daily living and live and work within the common environment. This means using everyday products and controls designed around the anthropometric measurements of the typical human upper-limb (Croney 1980). Therefore, it was considered that a first principles approach to a novel prosthesis would be to take reference from the human arm. Whilst a first principles approach appeared opportune in the light of emergent research investigating novel forms of actuation closer to biological models challenging previous prosthetic design constraints.

5.3.1.2 Case Study: Development of Analogous Forearm Joints

Introduction. The development of the forearm joints has been chosen as this starkly contrasts with conventional prosthetic approaches in providing the articulations needed for pronation and supination movements. Research into the articulations of the forearm followed from the first cycle of research – the production and evaluation of a tendon actuated model hand. Both quantitative and qualitative evaluation by end-users indicated that the majority of the articulations within the model were a close match to those of the human hand (Whiteley 2000). However, the articulations of the first model wrist were considered less successful. Although the wrist articulations gave a similar range of movement qualitative evaluation indicated that these did not provide a close mimic to the 'centres' of rotation of the human wrist. Additionally, although the use of computer controlled machining techniques had provided an appropriate means of producing the multiple joints required to construct the hand, from end-user evaluation, it appeared that their rectilinear form was not appropriate for a future prosthetic device (Whiteley 2000). Consequently, it was concluded that in pursuing further analogies of the more proximal joints, such as the forearm, articulations should provide as close a mimic as possible to those of the human limb whilst additionally endeavoring to maintain the form of the skeletal limb.

Existing components. Pronation and Supination movements of the forearm are important to effectively accomplish activities of daily living (Buckley et al. 1996). However, many prosthetic suspension systems for below elbow amputees use the epicondyles of the distal humerus (Figure 4a) to stabilize the prosthesis on the remaining limb (Figure 3a). In practice this provides stability to the 'hand' component and also a more reliable contact between the electrodes and the amputees skin, necessary for myoelectric control. However, it prevents any natural pronation or supination in the remaining limb. Consequently, either a passive or motorized component can be placed distal to the vestigial limb to replace this function. Figure 3b shows a typical 'wrist' for a body-powered device appropriate for a unilateral amputee. This comprises an axial strut connected to a circular plate with a concentric ring of holes operated by the amputee's contralateral hand. Figure 3c shows a powered wrist. The fitting of powered components is becoming more common (Martin 2000). However, problems still remain with their control (Datta and Brain 1992) and from the increased weight and bulk to the prosthesis (Martin 2000).

FIGURE 3: Existing components effecting pronation and supination function.

The Human Forearm. Rotation of the human forearm occurs by the rotation of the radius bone about the ulna bone (Smith et al. 1996). In a position with the arms by the side of the body and the palms of the hand facing forward, the forearm is supinated. If then the palm is rotated to face backwards the forearm is pronated. In the supine position the radius and ulna run parallel to one another, whilst in the pronated position the radius crosses the ulna (Kapandji 1982) (Figures 4a, 4b). These bones are cranked along their length to permit them to cross. This mechanical arrangement appears much more complex than that of the simple axle shown in Figure 3b. However, the grip of the human hand may be attributed to the comparatively large extrinsic muscles of the digits (Figure 4c) situated in the volume of the forearm (Chao et al. 1989; Kapit & Elson 1993). Therefore, the extrinsic finger tendons need to pass through the wrist. This passage would not be possible if the wrist simply rotated about a single axial strut (Kapandji 1982). Like the human arm the first model hand was tendon actuated, consequently, it was thought appropriate to follow the more complex human model in the analogy of the forearm articulations. Additionally, it was hypothesized that there may also be a cosmetic benefit to the amputee in achieving pronation and supination movements in this manner.

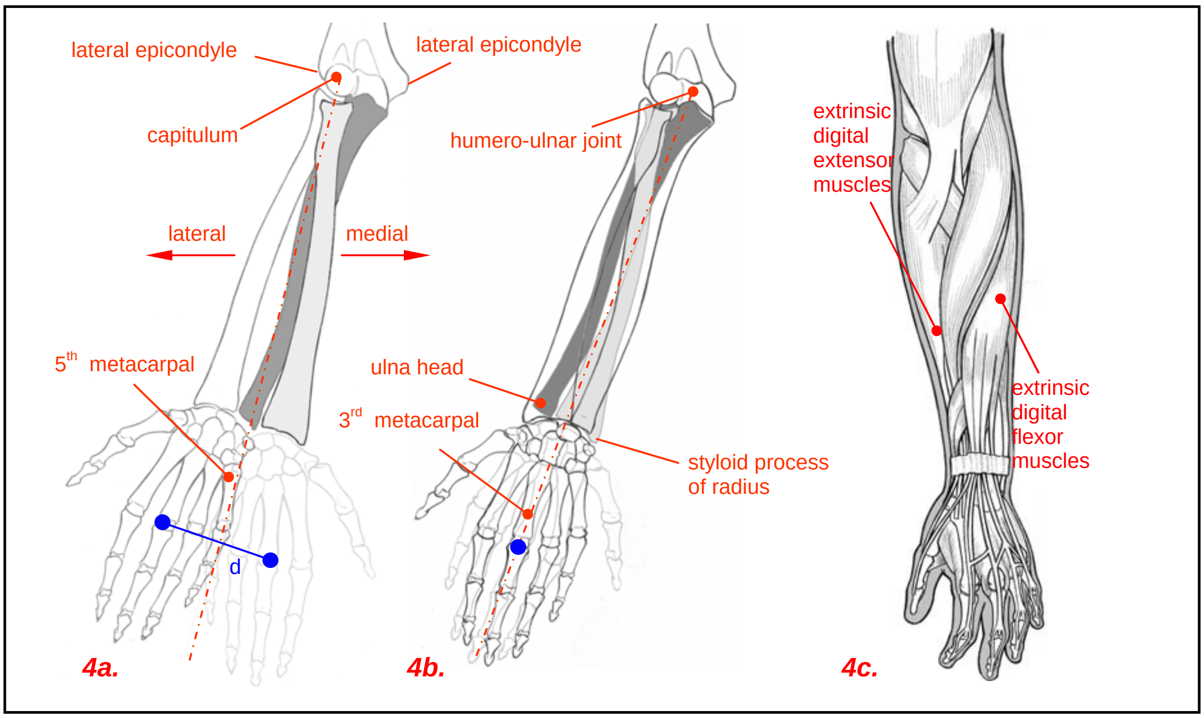


FIGURE 4: Diagrams to Explain Pronation and Supination in the Human Forearm.

Forearm Pronation and Supination. Biomechanics texts indicate that the rotation of the forearm occurs along an axis extending from the approximate centre of the capitulum to the distal head of the ulna (Norkin & Levangie 1992, Smith et al. 1996). It can be seen from Figure 4a that if this axis of rotation is extended it passes through the fifth metacarpal. Therefore, as the forearm is rotated, a point on the third metacarpal would move a distance 'd'. However, other anatomical texts refer to an 'axis of pronation-supination' (Kapandji 1982). This axis extends through the centre of the third metacarpal, Figure 4b, with the effect that the hand may be rotated without consequent translation of the third metacarpal (Amis 1990). For this to be correct requires that the distal ulna translates during forearm rotation, as shown in Figure 4b. In fact during forearm rotation the distal ulna appears to translate medially to laterally as the arm is pronated and reverses in direction as the arm is supinated. Kapandji 1982 indicates that this is an apparent translation of the distal ulna head and may be due to compensatory movements of humeral rotation. Kapandji 1982 states that without compensatory humeral rotation such a movement requires that either; there is an articulation between the distal humerus and proximal ulna (humero-ulnar joint, Figure 4b); or that the ulna is deforming. There is no radiographic evidence for distortion of the form of the ulna contributing to pronation-supination movements (Chao & Morrey 1978), therefore, debate has focussed on the existence of a humero-ulnar medial to lateral articulation. Researchers using LED's mechanically connected to cadaver limbs, report photographic evidence of distal ulna translations during pronation and supination movements (Youm et al 1979). Other research states the presence of medio-lateral rotation at the humero-ulnar joint providing 10 degrees of articulation during pronation and supination (Amis 1990). However, another group using radiographic techniques have detected no medio-lateral rotation at the humero-ulna joint (Chao and Morrey 1978).

Initial Measurements. Due to these uncertainties it was considered that further primary data were needed. Using a male subject with 50th percentile hand dimensions measurements of wrist and ulna head width were taken using digital vernier calipers. The wrist was palpated to find the styloid process of the radius (Figure 4b), and the position was correspondingly marked on the skin. This anatomical feature was chosen as it possesses minimal soft tissue coverage. The subject then positioned his arm onto a sheet of paper with a line ruled upon it. First the olercranon process (elbow bulge) was placed on this line, and then the arm was positioned so that the third finger was aligned with the same line. With the arm in pronated position, using an engineer's square, a mark was placed on the paper corresponding to medial position of the styloid process. This procedure was repeated with the arm in supine position. The distance perpendicular to the scribed line on the paper between the marks was then recorded using a steel rule accurate to 0.5mm. The position of the hand in pronation and supination was approximately parallel to the plane of the paper in transverse section, therefore it was reasoned that for the ulna to remain stationary during this movement, a value of (2*(wrist width)-(ulna width)) would be the expected distance recorded. During these initial experiments the wide error in the recorded results showed that it was difficult to ensure humeral rotation was not influencing the results. However, a difference of approximately 1/3 wrist width was recorded subtracting the recorded result from the theoretical result. This indicated that the mechanism of pronation-supination may be more complex than that of a single axis (Figure 4a). Consequently, closer investigation of the human forearm was necessary.

Methods. Drawing clear design principles appropriate for a prosthetic or robotic device from the human upper-limb is complicated by the subtleties of the human form, the variations between individuals, and that tissues of the human limb are often not as delineated as in artificial objects (Landsmeer 1976). Therefore, a research method was proposed that alongside a conventional literature review and quantitative analysis made extensive use of drawing as a primary means of increasing the researchers acuity of observation. Considering that good observation would provide a solid foundation upon which ideas for analogous joints could be formed. However, the recognition that drawings and designs are still abstractions was integral in this method and a stage of model making/prototype construction is included. This permits tangible models embodying these ideas to be assessed for their 'closeness' to the human limb both quantitatively and qualitatively by end-users. The end-users of this reported research are primarily amputees, however, prosthetists, occupational therapists and prosthetics manufacturers are also 'users' as they are affected by the design of prostheses.

Observational Drawing. Observational drawing studies were done on an A3 format. Details from these are included in Figure 5 to explain the reasoning process. Many of the observations made during the drawing process were in agreement with anatomical literature, however, certain observations pointed to subtleties of form not elucidated in the literature. For example, literature review indicated that the humero-ulnar joint possesses a single degree of rotational freedom for flexion and extension of the elbow (Norkin & Levangie 1992). Stating that this is due to the highly contiguous fit between trochlea ridge of the ulna within the trochlea groove (Kapandji 1982). However, the trochlea groove and trochlea ridge of the skeletal models were palpated and a large amount of freedom of movement was perceived. From observation of the form of the trochlea, it is evident that the groove is not sharp but akin to the depression around an hourglass (Norkin & Levangie 1992) (Figure 5a). Whilst the trochlea ridge of the ulna appears slightly sharper in its convexity than the trochlea groove is concave. This was initially attributed to the absence of cartilage in the skeletal joint. However, references to photographic cross-sectional studies indicate the trochlea and trochlea notch not to be totally contiguous on the lateral and medial borders, permitting the possibility of a medio-lateral rotation (Guyot 1990). Due to the observed differences in these forms it was reasoned that if a medio-lateral articulation exists it is likely to be close to the centre of the trochlea groove.

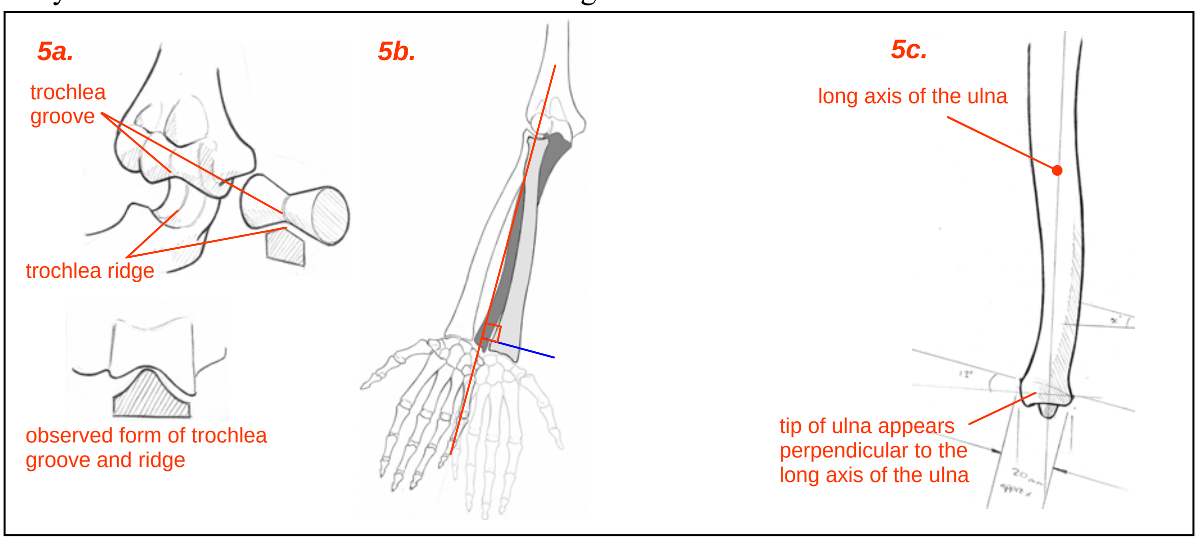


FIGURE 5: Details from observational studies of the proximal and distal forearm joints.

Observational drawings of the radius and ulna at the elbow (proximal) were followed by studies of the radius and ulna close to the hand (distal), as both proximal and distal radio-ulnar joints are stated as coupled during forearm rotation (Norkin & Levangie 1992). It was observed, and confirmed from the anatomical literature, that the distal radius possesses a concavity, which articulates against a cylindrical surface on the distal ulna (Kapandji 1982). From observational studies of this cylindrical surface, it did not appear to be orientated at an angle perpendicular to a line originating from the centre of the capitulum and extending to the centre of the head of the ulna (Figure 5b). Instead, the observed cylindrical surface appeared angled either, perpendicularly to the longitudinal axis of the ulna, or angled slightly proximally, medio-laterally

(Figure 5c). Further literature review indicated that the ligamentous structures between the radius and ulna are crucial in resisting translation of the radius relative to the ulna (Skahen et al. 1997). Although these could not be observed on the skeletal models, it was reasoned that the role of the most distal radio-ulnar ligament, the articular disc (Norkin & Levangie 1992), might be deduced from observation of its insertions and articulating surfaces. The articular disc inserts close to the ulnar styloid process and onto the medial and frontal aspect of the radius (Kapandji 1982). It was reasoned that if this 'disc' is considered inextensible then the approximately perpendicular rim of the ulna would guide the radius bone through a similar path, i.e. perpendicular to the long axis of the ulna. Additionally, it was noted that the concavity in the radius bone is larger in 'radius' (literal) than that the 'radius' (literal) of the ulna. Literature review indicated that radiographic research had identified this in the intact distal radio-ulnar joint (Cone et. al. 1983), indicating the possibility of further articulations at this interface.

Mathematical Model. The observational drawing previously undertaken was used as a basis for a two-dimensional trigonometric analysis to understand how the proposed joints would function when coupled together. During the observational drawing studies approximate measurements of anatomical features were recorded using a steel rule accurate to 0.5mm and annotated on the drawings. The distance from the centre of the trochlea notch to the distal centre of the ulna head (dashed) was recorded as 250mm; the distance between the centre of the capitulum and the trochlea groove (dotted) estimated at 22mm, and the radius of the cylindrical surface on the distal ulna (centrelined) estimated to a mean of 10mm. Two views of this set of links were drawn, one relating to the supine position where the centrelined link is to the right of the dashed link (Figures 6a, 6c), and a pronated view in which the centrelined link has been rotated 180 degrees perpendicularly to the dashed link to be to its left (Figures 6b, 6d). As the bones of the forearm are not understood to deform under normal pronation-supination movements (Chao & Morrey

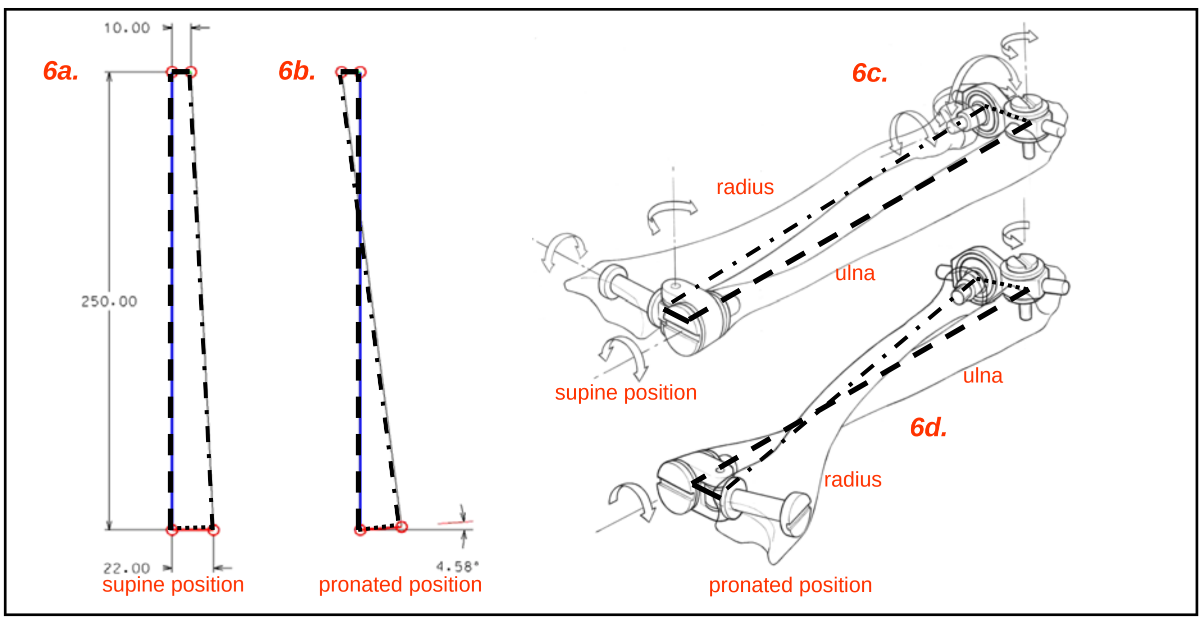


FIGURE 6: Forearm articulation link diagram.

1978) the links joining the articulations were considered of the same length in both cases. For ease of calculation the dashed link was considered stationary to the remaining links. Using simple trigonometry it was calculated that the dotted link would move through an angle of 4.6 degrees (counter clockwise) on full pronation. The dotted link represents the centre line on which ulna and radius rotations take place, however, these centres are collinear with the main axis of elbow flexion in the model. In the initial measurements it was this axis that had been chosen as the datum from which parallel translations of the distal ulna were measured. Therefore, considering the dotted link stationary a *clockwise* rotation of the ulna of 4.6 degrees represents a translation of Sin (4.58)*250mm = 20mm parallel to the dashed link. This was very close to the values recorded from the intact limb, and so the joint designs were detailed for prototype production.

Development of Joint Concepts: Distal Joint. Observational drawing indicated that movement of the articular disc against the distal ulna may act as a guide for the movement of the radius. Therefore the researchers designed a cylindrical track feature into the distal joint, comprising two large thrust washers to prevent uncontrolled distal or proximal movement of the radius relative to the ulna (Figure 7a). Observational drawings had highlighted the differences of curvature of the cylindrical surface of the distal ulna to the concave section of the distal radius; also relative movements between these surfaces have been reported during pronation and supination (Cone et al. 1983). Therefore, it was reasoned that if relative angular movement occurs, it would be at the interface of the two articulating surfaces. Consequently the design comprises a further two orthogonal articulations cross at a point at a distance equivalent to the radius of curvature of the distal ulna head (Figure 7a). Anatomical texts indicate that the range of movement of pronation and supination of the forearm is effected by, amongst other factors, the ulna styloid process (Kapandji 1982). An analogy of the ulna styloid is included as means of limiting the angular movement of the model joint. The proposed analogy in the model joint is a prominent fixing projecting into a semicircular track connected to the model ulna 'bone' (Figure 7a).

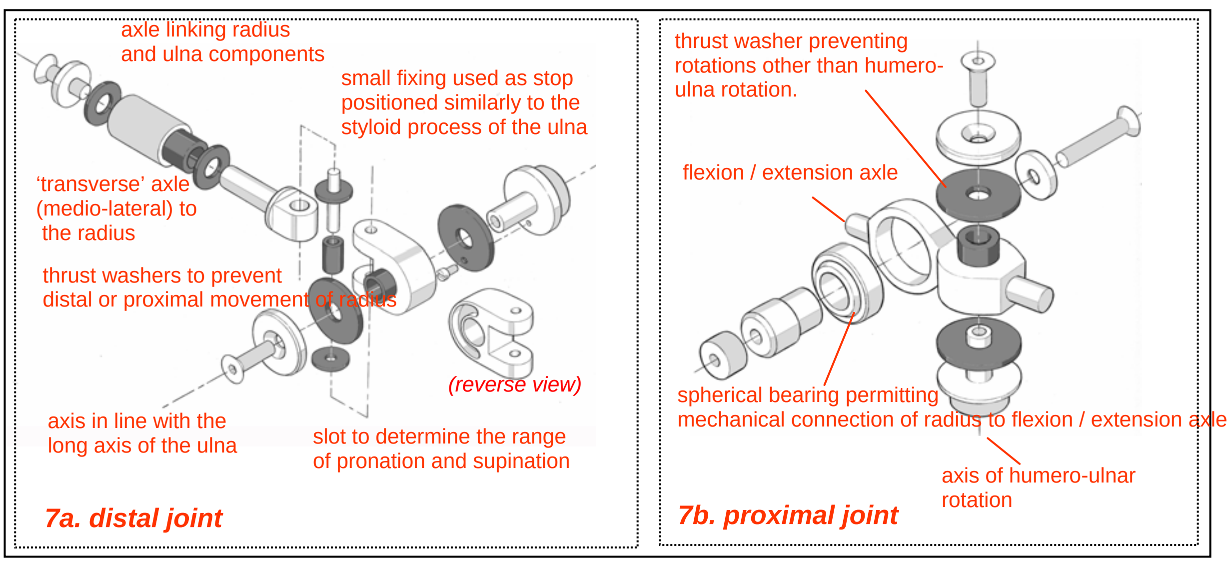


FIGURE 7: Exploded views of the proximal and distal forearm joints.

Development of Joint Concepts: Proximal Joint. The elbow flexes on a common axis that can be considered to run through the centre of the capitulum and the centre of the trochlea (Norkin & Levangie 1992). Therefore, an analogy was considered placing all the proximal articulations on a common flexion and extension axle (Figure 7b). It was reasoned that if the spherical bearing was connected to an axle common to the ulna, then it would only need a small range of spherical articulation to permit pronation and supination movements. Observational drawing indicated the position the radius against the capitulum to be approximately perpendicular to that of the connection of the ulna to the trochlea. Therefore, sketch ideas for the axle proposed that the axle be machined to allow the spherical bearing to be fitted at 90 degrees to the axis of medio-lateral rotation of the ulna (Figure 7b). The simple geometric forms of the proposed joints were conventionally machined from 'freecutting' stainless steel, whilst a polyamide polymer was chosen for bearings within the joints.

Evaluation: Joint Fixation Experiment. Once the joints were made, a model skeleton was used to make high definition silicone rubber moulds of the radius, ulna and humerus bones. Subsequently, a low contracting rigid polyurethane casting resin was poured into the mould cavities. The resin bones were then arranged in a posture with the ulna and radius flexed at approximately 90 degrees to the humerus, with the long axis of the ulna parallel to a scribed line on the mounting board. The radius and ulna were positioned in supine posture, whilst a two part polyester resin of a contrasting colour was applied between them. The humerus was then removed and the mounting board assembly taken to a vertical milling machine. The scribed line on the board was aligned with one of the axes of the milling machine. The centre of the trochlea notch was approximated on the resin cast, and this was taken as a datum. Using this datum 'pockets' were machined into the casts to insert the joints at positions corresponding to the calculated link lengths (Figure 6a, 6b). The joints were secured to the bones with more polyurethane resin. Subsequently, the resin binding the bones together was removed and the model was pronated and supinated. An indicator was placed on the styloid process of the model radius and a lightweight extension arm fixed to the flexion/extension axis of the proximal radio-ulnar joint. The model ulna remained firmly fixed to the mounting board with polyester resin. Measurements were then taken of the angular position of the proximal flexion/extension axle with respect to changes in angle of pronation and supination (Figure 8a).

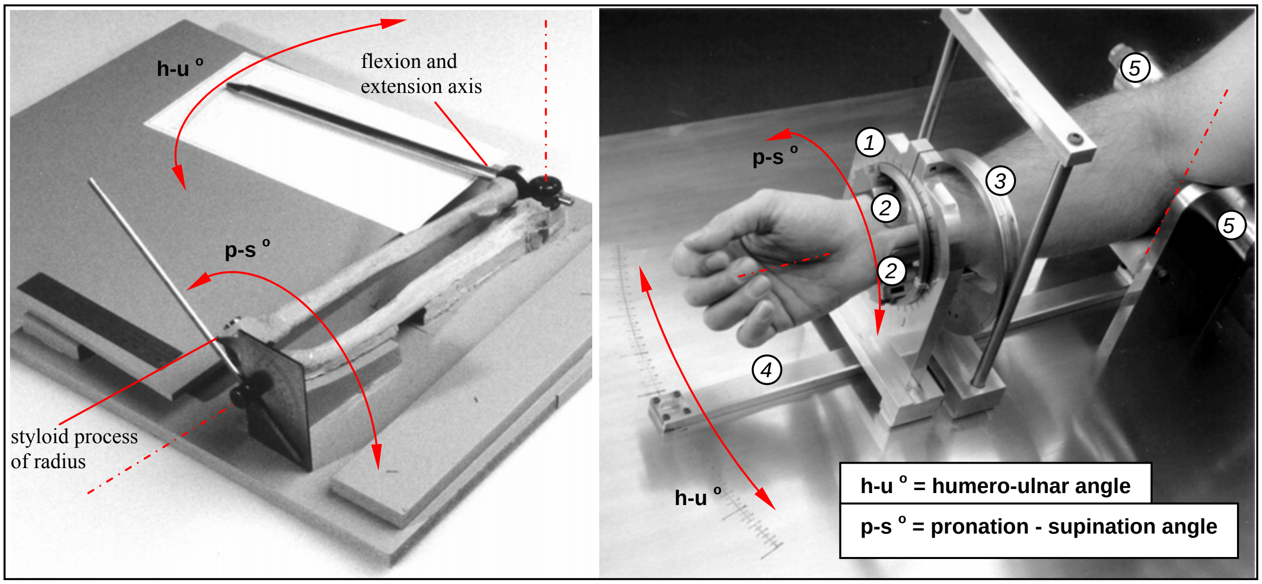


FIGURE 8: (a) Joint evaluation test rig (b) measuring splint.

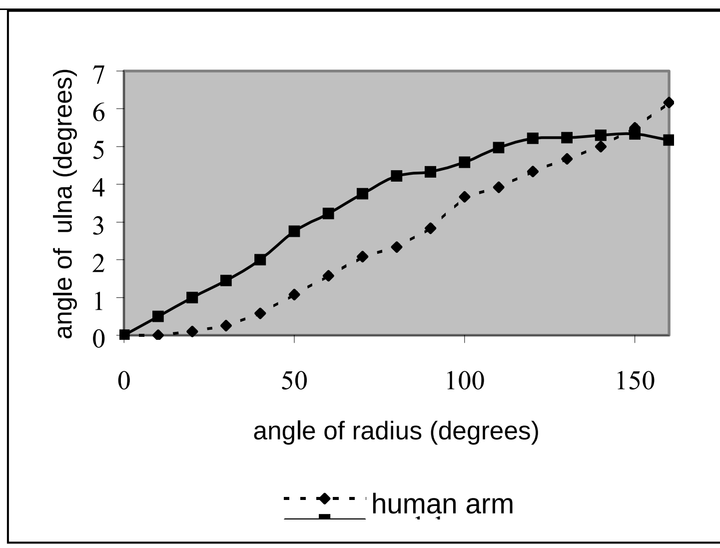
Evaluation: Pronation-Supination Splint. It was considered necessary to compare the resulting model pronation–supination movement with those of the intact human forearm. Previous studies on the pronation-supination movement of the human arm have been performed on cadaverous arms (Youm et. al. 1979). Cadavers have been used principally so that rigid markers can be affixed to the skeleton. More recent studies (Nakamura et al. 1994) have been performed using Magnetic Resonance Imaging (MRI). However, with these experiments it is unclear how the splints required to fix the arm within the imaging device constrain the movements of the arm. Conflicting results have been reported using radiographic methods: the research findings of (Cone et al 1983) report no evidence for a medio-lateral articulation at the humero-ulna joint whilst (Amis 1990) reports radio graphic evidence supporting a medio-lateral articulation up to 10 degrees.

To ascertain how closely the prototype joints were reproducing the movement of the ulna during forearm rotation it appeared appropriate to take measurements from an intact limb, where all the constraints of a mechanical splint were know. A male of normal build was chosen as the subject for these measurements. A close fitting splint was tailor made for the subject. This comprised a large plain annulus bearing (Figure 8b (detail 1)) with adjustable close fitting adjustable wrist clamps (detail 2). This was housed in a fixture with increments of 10 degrees marked upon it. The wrist clamp for the ulna was mechanically connected to a grooved hollow disc (detail 3). Using a vernier height gauge the disc was positioned so that the centre of the ulna was coincident with the centre of the disc (detail 3). The grooved disc rotated between two parallel ground bars connected to a stand with a ‘peg’ projecting from its base in line with the vertical centre line of the disc. This peg was slotted to the indicator bar (detail 4) to record angles of humero-ulnar articulation. The Proximal part of the forearm was secured in an elbow fixture with two adjustable pegs securing the humeral medial and lateral epicondyles (detail 5). Vertically beneath the elbow fixture and aligned with the approximate centre of the olercranon was a peg onto which the indicating bar rotated. Both the fixture for the wrist and grooved disc were manufactured with PTFE (polytetrafloroethylene) blocks at their base. The mounting board was made from polished aluminium plate, onto which a low surface tension lubricant was sprayed to further reduce friction between the mounting board and the distal clamping fixtures. The subject was firmly secured in the jig and the forearm rotated in 10-degree increments, whilst the angular position of the ulna was recorded from the indicator bar. The experiment was repeated three times, the mean values being recorded.

Results. Figure 9 shows the difference between the movement of the model ulna and that of a human ulna during pronation-supination movements. From these results it is evident that rotation of the bones was occurring before being indicated by the scale on the annulus bearing, due to movement of soft tissue within the wrist clamps. This was estimated by twisting the clamp whilst retaining the wrist in full supination. Although the clamps were tight around the wrist, it was found that this movement accounted for an initial estimated error of 10-20 degrees, due to the skin becoming taut before movement was recorded. All measurements were taken using 0 degrees (full supination) as the start position. Therefore, although currently there is a wide discrepancy between the angular positions in the mid-range, this can be slightly offset by soft tissue errors, if the maximum estimated error of 20 is used this brings the maximum discrepancy down to less than 1 degree, or less than 4mm of ulna translation.

Qualitative Evaluation by Lisa Halse, Osteopath. Alongside the quantitative evaluation of the joints the production of the prototypes permitted qualitative evaluation by experts in the movement of human joints. Before these evaluations took place the joints were placed into casts of the radius and ulna, similar to those shown in Figure 7a and the previous hand model fixed to it. The two professionals chosen to evaluate the joint designs were Lisa Halse, an osteopath in Sheffield, UK, and Professor John Stanley, an Orthopaedic Surgeon based in Manchester, UK. Lisa Halse was selected because she routinely uses palpation of intact limbs as part of diagnosis, and therefore was considered to have a keen sense of how normal joints feel.

	model	human
p-s °	h-u °	h-u °
0	0.00	0.00
10	0.50	0.00
20	1.00	0.10
30	1.45	0.25
40	2.00	0.58
50	2.75	1.08
60	3.22	1.58
70	3.75	2.08
80	4.22	2.34
90	4.33	2.83
100	4.58	3.67
110	4.97	3.92
120	5.22	4.34
130	5.30	4.67
140	5.33	5.00
150	4.83	5.50
160	4.53	6.17



p-s ° = pronation - supination angle, degrees
h-u ° = humero-ulnar angle, degrees

FIGURE 9: Graph to show relative humero-ulnar articulations of the model and an intact human arm.

The aims of the evaluation were for L. Halse to indicate from palpation where deviations from the original anatomy appeared to arise. During the evaluation L. Halse was encouraged to mark the model arm with adhesive paper markers where she considered the model diverted from the original anatomy. The interview took place in the Research Workshops at Sheffield Hallam University. Present were L. Halse, R. Erol, C. Rust and G. Whiteley. The evaluation lasted for approximately 45 minutes. Below are the salient points she made, together with quotes from the transcript of the interview.

Model Joint Movement: Distal Forearm Joint

'It's definitely mechanical…there's no sense of …things receding and coming back….It's <requires> a sort of springiness of a kind.'

Model Joint Movement: Proximal Forearm Joint

'…the kind of movement though, and feel of it,…it feels like its much more alive.'
'It's the form, because of the curve this way, and the curve that way, and it's rounded.'

Absence of Soft Tissue

L. Halse contrasted the model to the human forearm indicating the absence of soft tissue within the model was hampering her evaluation

'…you can feel if it's <original anatomy> alive. you can feel there's a sort of slight expansion, or rotation and contraction sort of feel to the tissue.'<considered absent in the model>

'…you've got to build on <to the model> some muscle..'
'..if you start putting on something elastic, you know something that can conform into that function <indicated contracted muscle of forearm> …and it has elasticity to go back again,..you can start building up soft tissue in the model it will feel more real..'
'It's brilliant, but it's nothing like a real hand because it's got no soft tissue'
'…I want it to have all the rest of the soft tissue on…, to feel'

Weight

On picking up the model for palpation L. Halse commented on the difference in weight between proximal and distal sections of the model.

'It's massively heavy <model elbow> … it won't integrate <with the movement of the amputees body> while it's got this massively heavy joint.'

The evaluation with Lisa Halse concluded with her requesting to review the model once again when analogies of the soft tissues had been devised for the model limb.

Qualitative Evaluation by Professor John Stanley, Orthopaedic Surgeon. The evaluation by Prof. J. Stanley took place at Northern General Hospital, Sheffield. Present were Professor J. Stanley, D. Stanley (fellow Orthopaedic surgeon), G. Whiteley and C. Rust. The evaluation lasted approximately 30 minutes, and was tape-recorded. Before Prof. Stanley was presented with the model arm to palpate the brief aims of the research were given. The aim of the evaluation was stated to inform the researcher where the model appeared close to the anatomy, and where it was viewed to divert from the anatomy.

Model Joint Movement

'As far as pronation and supination are concerned, that's fine <palpates the model> it doesn't actually work that way. But it's pretty close…it's actually reproducing the movement pretty accurately.'

'…in terms of the net effect, overall it's pretty good, in fact it's very good…you'd be hard pressed to tell the difference <between model and human forearm articulation>'

'…the axis of rotation <human forearm> is based on an interosseous membrane it doesn't work on two fixed linkages.'

Absence of Soft Tissue

'I'm trying <palpates model with eyes shut> to ignore what it looks like and try to see what it does.'

'..what you need to do is cover it in a rubber glove or something filled with silicone or saw dust, just to give it that damping effect…'

'…once you've <the researcher> got it powered up and some damping on it, it will be absolutely super.'

The evaluation finished with Professor Stanley indicating that he would like to evaluate the model wrist again when further work had been completed on the model's actuation.

Discussion. The quantitative results indicate that in the mid-range of the movement of the forearm there is a decrepancy between the type of movement generated by the linked joints and that generated by the intact human arm. Subsequent investigation has shown that this decrepancy can be minimized by slightly angling the cylindrical track of the distal unlar component relative to the long axis of the ulna. However, due to the errors encountered in the data gathered from the splint, it is considered that effort be focused in gaining further data from the intact arm before further revisions to the joint designs are made.

The qualitative reviews of the joints by experts have shown that there are difficulties with this method when there is not a soft tissue analogue as part of the model. However, these reviews have indicated the joints are reproducing a similar movement to that of the human forearm. These evaluations have also pointed to the real need for actuators that are more akin to skeletal muscle if prostheses are to appear more 'natural'.

Observational drawing has been used extensively in this research as a means of analyzing form and aiding the generation of analogous concepts. The production and evaluation of the forearm joints has shown that this method can yield useful results. It has also been demonstrated that in the analysis of the coupled distal and proximal radio-ulnar joints the use of multiple methods are valid when observational drawing and mathematical modeling are combined (Figure 6).

Future Work. Although this research has concentrated on achieving analogies of the articulations of the skeletal limb, some further work has started towards analogies at the

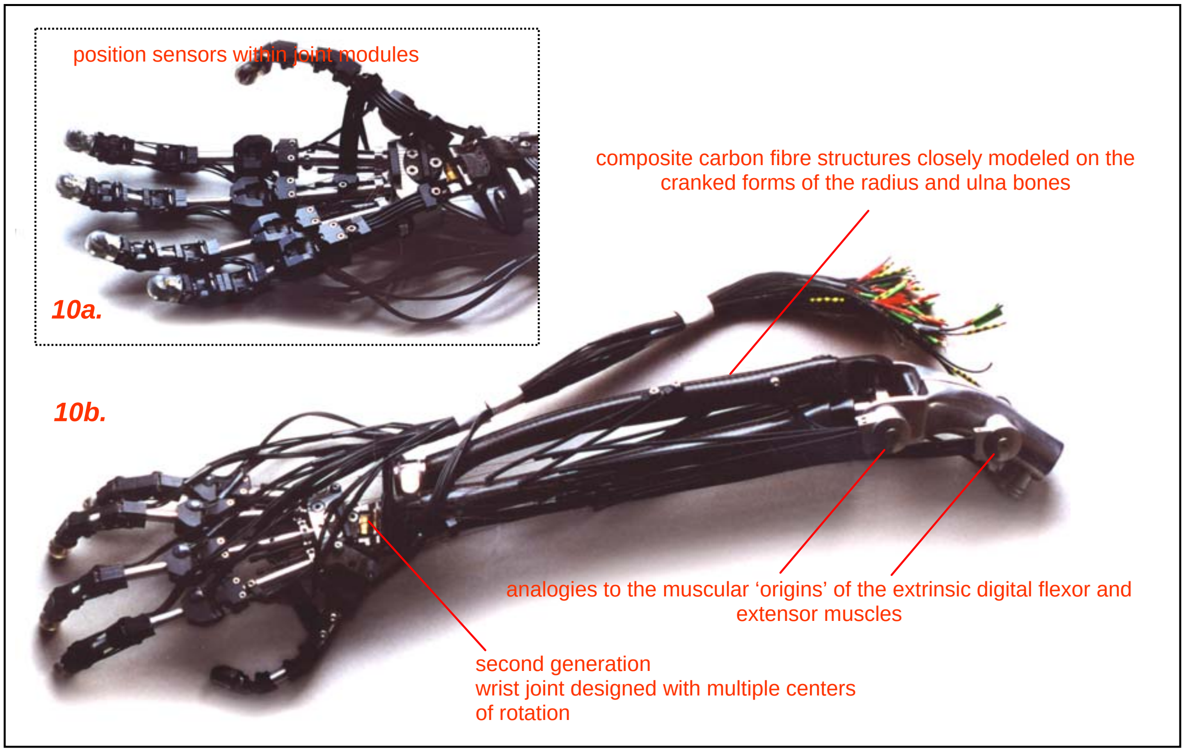


FIGURE 10: Current state of development of the anatomically analogous model limb.

muscular level looking towards appropriate actuators and sensors. In these investigations an arm prosthesis has been developed, shown in Figure 10, complete with position sensors within the finger joint modules. From the largely successful results of the skeletal level research it is considered that such 'biomimetic' methods may be of value in tackling the further challenges of actuation and control.

5.3.2 Biomimetic Actuation

Critical to the advancement of biomimetic O&P appendages is the development of an actuator that behaves like muscle. In this section we first review conventional actuator technologies currently employed in the O&P field. We then discuss current developments in biomimetic actuator research that may one day prove critical to O&P technology. A series-elastic actuator is presented in which a tendon-like spring is placed in series with a motor for accurate control of output force. We also present synthetic and tissue-based muscle actuators, including descriptions of both polymer-based and actin-myosin force-producing strategies.

5.3.2.1 Conventional O&P Actuators

Along with limitations in current control methodology, actuation also presents constraints to novel designs in the O&P field. Hydraulic and pneumatic actuators have been experimented with in upper extremity prostheses (Kinnier 1965, Sheridan & Mann 1978). However, both technologies require comparatively bulky control valves if they are to be linked to electrical control, and the pneumatic approach often produces undesirable exhaust noise. Consequently, DC electrical motors are considered the most practical solution for upper extremity prostheses (Kyberd 1990) but these actuators are not ideal in providing the comparatively high torque, low speed movements necessary in a prosthesis, since a bulky, and often heavy gear reduction is required.

5.3.2.2 Series-Elastic Actuators

Recently a series-elastic actuator (Pratt & Williamson 1995) was developed for humanoid walking robots that may be applicable to the O&P field. The device has a tendon-like spring positioned in series with an electric motor where output force is equal to the position difference across the series elasticity multiplied by its spring constant. By controlling the deflection of the series spring, output force, rather than position, is controlled. Since position is easier to control accurately across a gear train compared to force, the series spring lowers the force errors typically caused by gear trains. Additional advantages of the series spring are to low-pass filter shock loads and to provide for the possibility of energy storage in each actuator module. In the biomimetic control section of this chapter, an ankle-foot orthoses is described in which a series-elastic actuator modulates joint stiffness for the treatment of drop-foot, a gait pathology resulting from stroke, cerebral palsy, multiple sclerosis, or traumatic injury (Blaya 2002).

5.3.2.3 EAP Actuators

Developments utilizing EAP (Electroactive Polymers), i.e. materials that can directly transduce electrical energy to mechanical work, may be appropriate for a future generation of O&P devices (Baughman 1996, DeRossi et al. 1992, Lawrence et al. 1993). In order to be electrically stimulated the first generation EAP materials (polyelectrolyte gels) needed to be between two separate electrodes in an electrolyte bath; however, the second-generation materials can be used as electrodes themselves (Baughman et al. 1990; Otero & Sansinena 1997) since they are themselves electrically conducting. When used as electrodes conducting polymers have been shown to react much more rapidly than polyelectrolye gels - in the order of seconds rather than 10’s of seconds (Della Santa et al 1997). However, the strains generated are smaller than those of first generation materials at between 0.5 and 10 percent (Baughman et al 1990; Della Santa et al 1997), whilst the force per unit area (stress) produced is comparatively large (Smela et al 1995). Consequently, mechanical configurations such as the unimorph have been utilized that can amplify these small strains and make use of their potentially high work capacity (Della Santa et al 1997).

Conducting polymers appear particularly appropriate to O&P systems as their action is silent, and once in the given position no further energy is needed to maintain position (Kaneto & MacDiarmid 1995). Recent research has utilized these materials in conformations ‘biomimicking’ the sarcomere of skeletal muscle (Jung et al. 2001). In this manner, the bulk characteristics of actuation system would be created from the combined contributions of many ‘linear stepper-motor’ type mechanisms. Through characterization of the many EAP materials now developed and in combination with an appreciation of scaling effects (Bar-Cohen et al. 2001) it appears that such an approach may result in an actuation scheme highly appropriate to O&P devices.

5.3.2.4 Muscle Tissue Actuators

Although a great deal of research has been conducted to develop an actuator technology with muscle-like properties, engineering science has not yet produced a motor system that can mimic the contractility, energetics, scalability and plasticity of muscle tissue (Hunter et al. 1991). Recent investigations (Dennis & Herr 2000) have begun to examine the feasibility of using *animal-derived* muscle as an actuator for artificial devices. To develop robust and controllable muscle-powered systems, formidable technical hurdles must be overcome. Perhaps researchers in the past did not consider muscle a viable actuator because of tissue robustness problems. Critical to improving the fatigue-life of muscle tissue is to advance electromechanical, chemical, and genetic intervention technologies designed to maintain muscle contractility and mass *in vitro*. Although preliminary research has been conducted in this area (Harris & Miledi 1972, McDonagh 1984, Dennis 1998), considerable work still remains.

Another area of difficulty when using a living tissue actuator is control. To examine the controllability of muscle tissue, a hybrid robotic fish (Figure 11) was recently constructed comprising a synthetic elastomer tail actuated by a single pair of whole muscle explants from frog semitendinosus muscle (Dennis & Herr 2000). In this device, a microprocessor controls muscle contractions by modulating electrical signals to each muscle actuator across two electrodes near the myotendonous junction. During performance evaluations, the robot swam through an amphibian ringer's solution supplemented with 2-g/L glucose, a broad-spectrum of antibiotic and antimycotic supplements, and aerated with non-filtered room air. The robot operated intermittingly for 7 hours. During this time, the machine performed basic swimming maneuvers such as starting, stopping, turning and straight-line swimming at speeds approaching 0.5 body lengths per second, suggesting that controlling hybrid devices might be an achievable goal.

Although difficulties remain, tissue actuators would offer certain advantages over synthetic actuator strategies. Tissues would generate force quietly, allowing O&P devices to operate without detection. They might also be adaptive, responding to varying work loads by modulating their structure to meet specific task demands. Still further, tissue actuators would offer an improved transduction efficiency and mechanical power output. Working aerobically, muscle can generate up to ~4000kJ of work from just 1 Kg of glucose (Wilkie 1960, Woledge et al. 1985). And for its size, muscle can generate a great deal of power (~200 Watts/Kg, Woledge et al. 1985), enabling O&P appendages to be lightweight but still powerful.

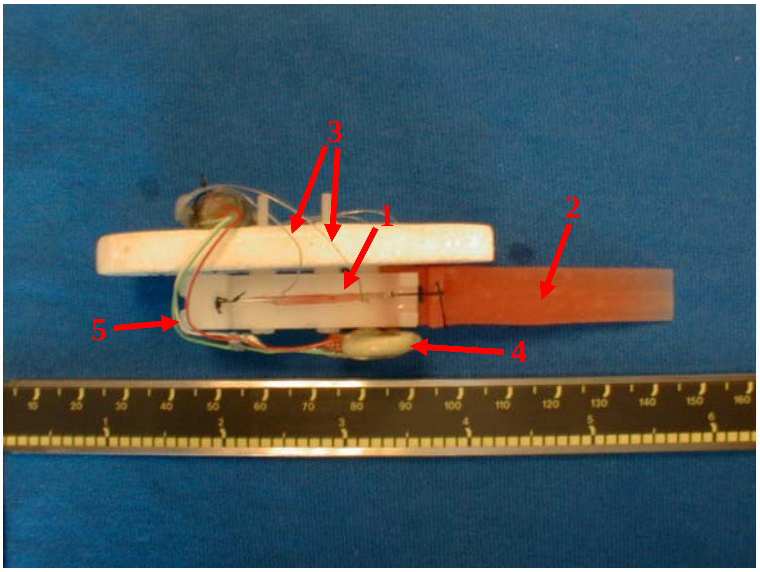


FIGURE 11: A side view of the hybrid robotic fish. To power robotic swimming, two frog semitendinosus muscles (1), attached to either side of elastomeric tail (2), alternately contract to

move the tail back and forth through a surrounding fluid medium. Two electrodes per muscle (3), attached near the myotendonous junction, are used to stimulate the tissues and to elicit contractions. To depolarize the muscle actuators, two lithium ion batteries (4) are attached to the robot's frame (5). During performance evaluations, the robot swam through a glucose-filled ringer solution to fuel muscle contractions.

5.3.3 Biomimetic Control

In the context of O&P devices, an important metric for system control is whether the resulting dynamics are biologically realistic. The scientific investigation into the development of biomimetic control strategies largely began in 1962 when Tomović and Boni developed the Belgrade hand, a device that resembles a human hand in both structure and movement (Tomović & Boni 1962). Following their seminal work, investigators advanced biomimetic devices in many fields of study, including arm and leg robots and O&P devices for upper and lower extremity amputees (Jacobsen et al. 1982, Popovic & Schwirtlich 1988, James et al. 1990, Popović et al. 1991). Although important strides have been made in the advancement of human-machine systems, complete restoration of natural movement is difficult even today and is quite often not achieved due to limitations in actuator design and control technique (Schaal 1999; Popovic & Sinkjaer 2000; Pratt 2000).

In this section, we review research areas critical to the advancement of biomimetic control strategies. Examples of biomechanical models of human movement are described and the potential importance of such models in the development of O&P control systems. Also described are locally controlled O&P joints that employ only mechanical sensing as control inputs, the use of noninvasive EMG signals to predict limb biomechanics, and hybrid sensory architectures that use both *in vivo* muscle force and external mechanical sensory information.

5.3.3.1 Human Movement Biomechanics

For O&P devices to move with a high level of stability and biological realism, biomimetic control strategies must be advanced that are consistent with biomechanical principles of human movement. In this section, two examples of biomechanical models are presented that enhance our understanding of human movement both from a joint level and from a global center-of-mass perspective. For each model, we discuss the potential importance of the biomechanical principles to the advancement of O&P control systems.

Shape and Roll of the Human Leg. Recently Hansen (2002) examined the roll over shape of the human foot, the roll-over shape of the human ankle/foot complex, and the roll-over shape of the human knee/ankle/foot complex during walking. Roll-over shape is the effective rocker geometry that the human locomotor system conforms to during the stance phase of walking. As an example, the roll-over shape for the knee/ankle/foot complex during walking is found by transforming the location of the center of pressure (CP) into a coordinate system found by placing markers at the ankle and the hip (the ankle marker is the origin and the markers are assumed to be in the sagittal plane). For simplicity, this knee/ankle/foot roll-over shape can be approximated by a circle, which is a shape that one can roll over on.

Hansen et al. (2002) found that the radius of the roll-over shape is approximately one third of the length of the leg, and that the roll-over shape (radius of curvature and orientation) is nearly invariant with walking speed. They also observed that prosthetic feet providing good function tended to have roll-over shapes similar to the roll-over shape of the human knee/ankle/foot system. Artificial foot shape should have arc lengths, arc radii, and arc orientation similar to the human systems, and these values, if possible, should remain reasonably constant with walking velocity.

It appears that it may be possible to use the roll-over shape of human systems as a scientific way to design and evaluate artificial feet, and to computationally align an artificial foot to the more proximal aspects of an artificial limb. Since human shape orientation is modified when walking up or down ramps (Hansen et al. 2002), the roll over shape concept may be the basis for an intelligent active system that uses the desired shape as a control target. Under microprocessor control, the limb would adapt its alignment to both level and inclined surfaces to effectively match the human roll-over shape of the knee/ankle/foot complex.

Conservation of Angular Momentum. Biomechanical investigations (Popovic & Herr 2002) have determined that a large class of human movements, including standing and walking, support conservation of total angular momentum $\vec{L}$ about the body's center-of-mass (CM), or

$$\frac{d\vec{L}}{dt}\Big|_{CM} = 0 . \tag{1}$$

Angular momentum is a conserved physical quantity for isolated systems where no external moments act on the body's center-of-mass. However, in the case of human locomotion, where the body interacts with the environment (ground reaction forces), there is no ***a priori*** reason for this relationship to hold. The constancy of angular momentum is not a necessary condition for stability in legged systems. Three dimensional passive walking toys, for example, do not conserve angular momentum in the CM frame. For these systems, angular momentum is periodic but is not conserved throughout time increments considerably smaller than the movement period. It is asserted here that angular momentum is locally conserved *throughout a movement cycle*.

A requirement of the constancy of angular momentum is that the sum of torques about the CM is equal to zero. This condition gives a simple relationship between the center-of-pressure CP, center-of-mass CM, and the CM force vector $\vec{F}$. A simple derivation gives

$$\vec{F}_{hor} = k\left(\vec{r}_{CM}\big|_{hor} - \vec{r}_{CP}\right) \tag{2}$$

where $k = F_{ver} / z_{CM}$ is a global stiffness, and *hor* and *ver* are abbreviations for the horizontal and vertical vector components, respectively.

In Figure 12, the CM and CP positions, $\vec{r}_{CM}$ and $\vec{r}_{CP}$, together with global vertical stiffness, k , are used to predict the horizontal forces that act on the body's center-of-mass in human standing and walking. The model (dotted lines) is in good agreement with the force plate data (solid lines). Preliminary results with advanced techniques for CM position calculation show agreement on the 95% level. Biomechanical studies suggest that this conservation principle is broadly applicable to humans, and may lend considerable insight into how future O&P systems might be controlled to achieve a high level of biological realism and stability.

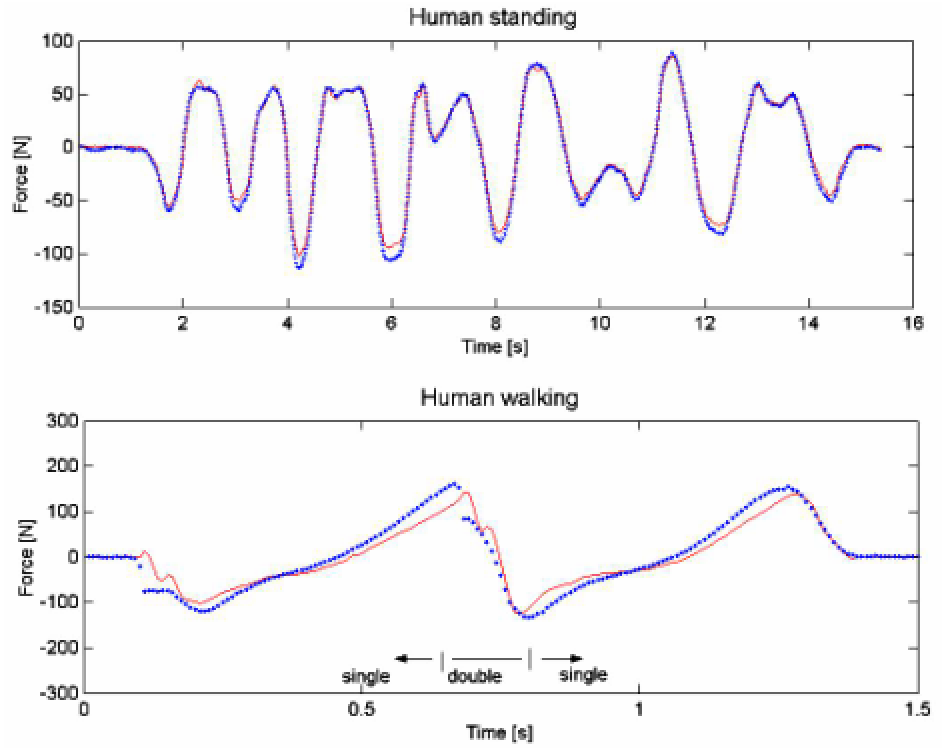


FIGURE 12: For human standing and walking, horizontal ground reaction forces measured from a force plate are compared to horizontal forces predicted from the conservation angular momentum model. The upper plot shows the medial-lateral center-of-mass forces in standing, and the lower plot, the anterior-posterior center-of-mass forces during level ground walking. Solid lines are force plate data and dotted lines are model predictions.

Summary. Human biomechanical models not only enhance our understanding of the way we move but also motivate control methodologies for anthropomorphic devices. In the next section, biomechanical models guide the control systems of autonomous microprocessor-controlled O&P leg devices designed to improve gait function.

5.3.3.2 Case Study: Autonomous locally Controlled O&P Leg Systems

Introduction. In order for individuals suffering from leg dysfunction to walk in a variety of circumstances, leg rehabilitation devices must provide stance control to limit buckling when weight is applied to the device. In addition, leg devices must also provide swing phase control so that biologically realistic dynamics emerge during swing. Unlike a biological leg, an autonomous leg orthosis or prosthesis, using only local mechanical sensing, must accomplish both stance and swing control without direct knowledge of its user's intent or of the environment. Rather, such a device must infer whether its user desires stance or swing behavior and predict when future stance/swing transitions should occur. Such a device must also determine when dramatic changes occur in the environment, as for example, when an amputee decides to lift a suitcase or walk down a slope. An autonomous leg device must not only be safe to use, but should also help the patient walk in a smooth and non-pathological manner. Conventional prosthetic knees often force the amputee to walk with an awkward gait. As an example, many prosthetic knees lock up throughout early stance, not allowing the amputee to go through normal knee flexion and extension motions typically observed in normal gait (Gard 1999). The amputee is therefore forced to roll over a perfectly straight leg, resulting in large vertical fluctuations in the amputee's center-of-mass and diminished shock absorption.

In this section, two microprocessor controlled, autonomous devices for leg rehabilitation are presented: 1) an external knee prosthesis for trans-femoral amputees; and 2) a force-controllable ankle-foot orthosis to assist individuals suffering from drop-foot, a gait pathology resulting from muscle weakness in ankle dorsiflexors. In these devices, muscle-like actuators and biomimetic control schemes are employed to enhance patient stability, speed and dynamic cosmesis. Patient-adaptive control schemes are discussed in which joint impedance is automatically modulated to match patient-specific gait requirements. By measuring the total time that the prosthetic foot remains in contact with the ground in walking, the prosthetic knee controller estimates forward speed and modulates swing phase flexion and extension damping profiles to achieve biological lower-limb dynamics. For the ankle-foot orthosis, joint stiffness is automatically adapted to permit a smooth and biological heel strike to forefoot walking transition in drop-foot patients. Using only local sensing and computation, the adaptation schemes presented here automatically modulate joint impedance throughout the stance and swing phases of walking. The control schemes are shown to enhance some features of above-knee amputee and drop-foot gait. For both the orthotic and prosthetic devices, mechanism design, control and device functionality are addressed.

Methods: Patient-Adaptive Knee Prosthesis. Using state-of-the-art prosthetic knee technology, a prosthetist must pre-program knee damping levels until a knee is comfortable, moves naturally, and is safe (Dietl & Bargehr 1997, Kastner et al. 1998, James et al. 1990). However, these adjustments are not guided by biological gait data, and therefore, knee damping may not be set to ideal values, resulting in the possibility of undesirable gait movements. Still further, in such a system, knee damping levels may not adapt properly in response to environmental disturbances. In this study, an external knee prosthesis is presented that automatically adapts knee damping values to match the amputee's gait requirements, accounting for variations in both forward speed and body size (Wilkenfeld 2000, Deffenbaugh et al. 2001, Herr et al. 2001). With this technology, knee damping is modulated about a single rotary axis using magnetorheological (MR) fluid in the shear mode, and only local mechanical sensing of axial force, sagittal plane torque, and knee position are employed as control inputs (See Figure 13). With every step, the controller, using axial force information, automatically adjusts early stance damping. When an amputee lifts a suitcase or carries a backpack, damping levels are increased to compensate for the added load on the prosthesis. With measurements of foot contact time, the controller also estimates forward speed and modulates swing phase flexion and extension damping profiles to achieve biologically realistic lower-limb dynamics. For a normal walking cycle, the maximal flexion angle of the knee during the swing phase falls within a narrow angular range, approximately 60 to 80 degrees for moderate to fast walking speeds (Wilkenfeld 2000). The knee controller automatically adjusts the knee damping levels until the swinging leg falls within the biological angle range for each foot contact time or forward walking speed.

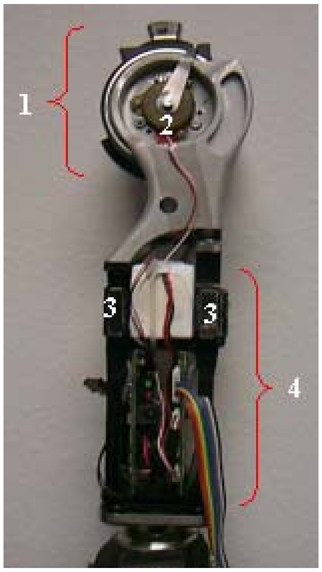


FIGURE 13: An external knee prosthesis for transfemoral amputees. The damping of the knee joint is modulated from step to step to control the movement of the prosthesis throughout each walking cycle. The prosthesis comprises magnetorheological brake (1), potentiometer angle sensor (2), force sensors (3), and battery and electronic board (4).

Methods: Patient-Adaptive Ankle-Foot Orthosis. A powered ankle-foot orthoses is presented for the treatment of drop-foot, a gait pathology most commonly caused by stroke, cerebral palsy, multiple sclerosis or traumatic injury. Drop-foot results from a particular muscle impairment in the anterior compartment of the leg where a patient is unable to dorsiflex the ankle or lift the foot. For walking, the major complications of drop-foot are 1) slapping of the forefoot after heel strike and 2) dragging of the toes at the beginning of each swing phase. The ankle-foot orthoses, shown in Figure 14, employs a force-controllable actuator and control algorithms based on biomechanical models of normal ankle function (Blaya 2002). Attached posteriorly to the ankle-foot orthosis is an actuator comprising a spring placed in series with an electric motor like a tendon in series with a muscle (Pratt & Williamson 1995). This series elasticity enables the system controller to modulate force instead of position (See **Series-Elastic Actuators**). For this spring plus motor system, output force is proportional to the position difference across the series elasticity multiplied by the spring constant. By applying a position control on the spring, force or torque can be controlled across the orthotic joint.

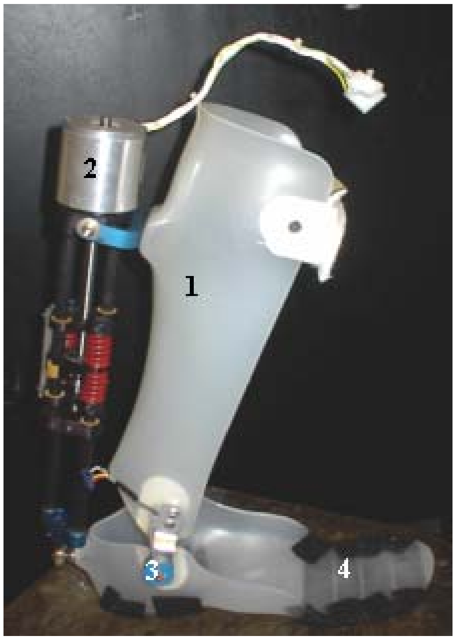


FIGURE 14: An actuated ankle-foot orthoses. This device is designed to treat drop-foot, a gait pathology resulting from stroke, cerebral palsy, multiple sclerosis, or trauma. Using the actuator, the stiffness of the ankle joint is modulated from step to step to control the movement of the foot during controlled plantarflexion. The ankle-foot orthoses (1) comprises a series-elastic actuator (2), potentiometer angle sensor (3), and capacitive force sensors (4).

The external sensors on the orthosis measure ankle joint position (potentiometer 3) and applied forefoot force (capacitive force sensors 4). These sensory data are then used to control 1) joint stiffness during the controlled plantarflexion phase of walking, and 2) ankle dorsiflexion angle at the start of the swing phase (to alleviate the toe dragging complication of drop-foot). The control law for controlled plantarflexion is a linear spring law where the actuator applies a torque proportional to joint position. As is shown in Figure 15, this linear control is consistent with the response of a normal human ankle during level ground walking (Palmer 2002). Although a normal ankle exhibits a spring-like response, the spring does not behave passively but seems to be actively controlled by the body's nervous system, changing stiffness from step-to-step even during steady walking. Hence, the stiffness of the applied virtual spring on the orthosis is not constant but is adjusted by the controller to optimize the heel strike to forefoot transition. Characterizing drop-foot is a high frequency force spike measured by capacitive force sensor 4 at the moment when the forefoot strikes the ground. The orthosis controller increases controlled plantarflexion stiffness until the high frequency force spike denoting forefoot collision is minimized.

Results. In Figure 16, the knee controller is shown to successfully adjust knee damping such that the swing phase peak flexion angle falls within an acceptable biological range of 60 to 80 degrees for moderate to fast walking speeds. Ovarall we find the adaptation scheme successfully controls early stance damping and swing phase extension damping, enabling amputees to flex

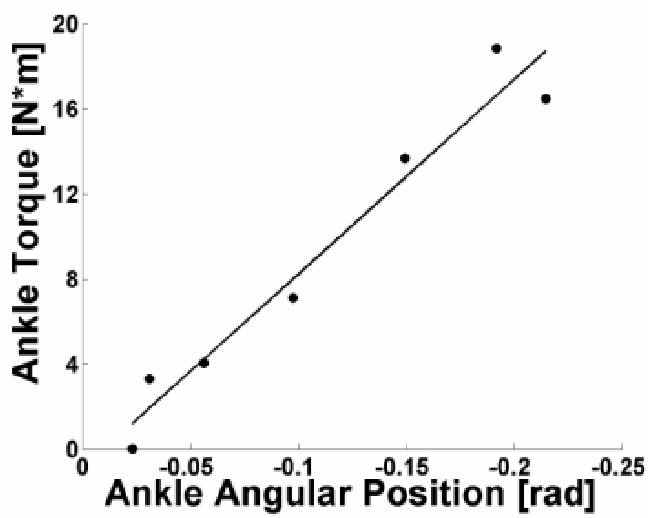


FIGURE 15: Ankle torque versus position data for a normal subject walking on a horizontal surface. Only ankle data during the controlled plantarflexion phase of walking are shown. Although data for just one subject and one walking step are plotted, the human ankle behaves like a linear spring throughout early stance independent of walking speed, storing and releasing energy and allowing for a smooth heel to forefoot strike sequence (Palmer 2002).

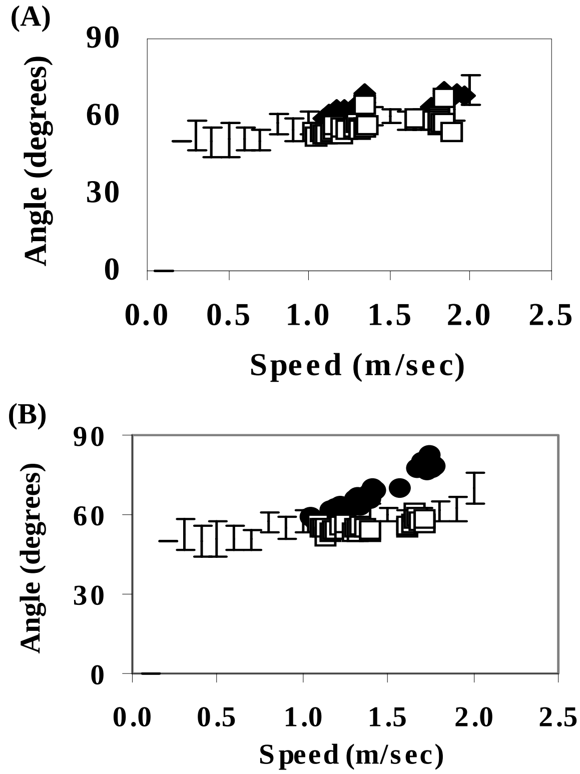


FIGURE 16: Peak flexion angle during the swing phase versus walking speed for one subject using the adaptive knee (plot A, filled triangles) and a non-adaptive, mechanical knee (plot B, filled circles). In (A) and (B), the subject's sound side leg is shown (open squares), along with reference data from unimpaired walkers of comparable height and body size (error bars). In (A), peak flexion angle is consistent with biological data, but in (B) the peak angle increases with increasing speed.

and extend throughout early stance and diminishing swing leg accelerations when the knee prosthesis reaches full extension just prior to heel strike (Wilkenfeld 2000).

For the orthotic device, we find that when the applied ankle stiffness is too low, excessive forefoot collisions occur, causing the drop-foot condition. To alleviate this complication, the orthotic controller increases stiffness at each walking speed and for each terrain until forefoot collision forces are minimized. Given this biomimetic control, we find the system provides a more biological and stable ankle response for drop-foot patients (Blaya 2002). Without an orthotic assistive device, unilateral drop-foot patients typically walk with an apparent gait asymmetry in which the step length on the affected side is smaller than on the unaffected side. We find that the variable stiffness ankle-foot orthosis improves gait symmetry by increasing step length and step time on the affected side.

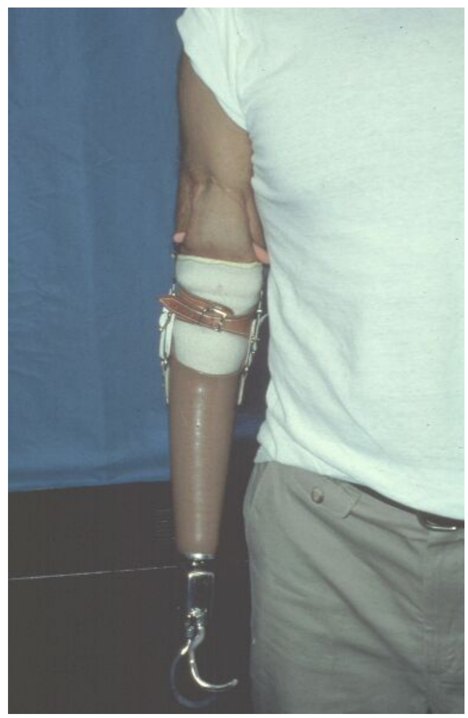
Summary and Future Work. In this section we report on autonomous leg devices that employ local mechanical sensors and computation to improve the locomotory capability of individuals suffering from gait pathology. Although some improvements in gait were observed, a great deal of work still remains. The fact that only local mechanical sensors were imployed in the O&P devices led to dramatic limitations in the systems' ability to assess user intent. Such devices cannot determine whether a patient wishes to turn to the right or to the left, or that an obstacle falls directly in a patient's intended pathway. Thus, an important area of research is to develop distributed sensory systems that can better assess the intent of the user.

5.3.3.3 Distributed Sensing

In this section we describe how invasive measures of muscle force, and noninvasive measures of EMG activity, can be used to assess human movement intent. We first describe upper extremity prostheses in which muscle tissue directly modulates prosthesis gripping force. We then discuss how noninvasive EMG measures can be used to foresee the future biomechanics of a human, and might one day be employed in O&P control strategies to predict future limb movements desired by the user.

Cineplasty. Around 1900, Italian surgeons investigated the possibility of bringing muscle forces of an amputated limb outside the body. They were not successful clinically with their efforts. However, in 1915 the German surgeon Ferdinand Sauerbruch, who was not aware of the Italian work, was able to carry out this procedure successfully on persons with limb loss. The procedure, called muscle tunnel cineplasty, was used widely in Germany during and after WWI by below-elbow amputees to control their prostheses. After WWII it became a popular procedure in America.

FIGURE 17: The cineplasty technique. Here the biceps muscle is released from its insertion so that it can move freely without flexing the limb at the elbow, and a hole is fashioned through the muscle and lined with skin. If a pin is placed through the muscle's tunnel and attached to a control cable, the biceps muscle can exert external forces to close a prehensor device so that the pinch force is proportional to the muscle force.



In biceps tunnel cineplasty, as shown in Figure 17, the biceps muscle is released from its insertion so that it can move freely without flexing the limb at the elbow. A hole is fashioned through the muscle and lined with skin. With practice, the user learns to contract the biceps without contracting other elbow flexors such as the brachialis and vice versa. If a pin is placed through the muscle's tunnel and attached to a control cable, the biceps muscle can exert external forces in excess of 250 N. This force can be used in a prosthesis to close a prehensor device so that the pinch force is proportional to the muscle force. The control method provides sensory feedback through the skin, muscle, and other tissues so that the user has a natural sense of force exerted, movement velocity, and tunnel position. It is a surgically created system that has distributed sensing. The system has excellent dynamic response. Figure 17 shows the Lebsche version of the Sauerbruch procedure, which uses only one control source. Sauerbruch preferred two control sources, one from an agonist muscle and one from its antagonist.

Muscle tunnel cineplasty, and versions of it, have the potential to be used advantageously in many ways to control artificial arms (Weir et al. 2001). The procedure has fallen out of favor with surgeons and others largely for esthetic reasons. However, the distributed sensing and the direct connection between the muscle and the object being controlled often results in exquisite control.

Case Study: EMG Signals Anticipate Human Movement. Recent investigations (Hoffman & Herr 2002) support the hypothesis that EMG peripheral neural signals can be used to anticipate human movements approximately 100ms in advance. These results are important because they suggest that EMG sensory data could be used to foresee the future biomechanics of a human, thereby making it possible to anticipate the movement intent of the O&P user.

Methods: EMG data analysis techniques. A number of experiments involving basic balancing tasks were performed for the purpose of collecting simultaneous position, force, and EMG data. Tasks included lateral swaying motions in standing double support, at different speeds, with both sinusoidal and random patterns analyzed. Position trajectories were recovered using a Vicon motion capture system (Vicon Motion Systems), and force trajectories using force plates (AMTI OR6-5). EMG signals were collected for the following leg muscles: Iliacus, Gluteus Medius, Adductor Longus, Vastus Lateralis, Rectus Femoris, Gastrocnemius, Soleus, and Tibialis Anterior.

Methods: Neural Net Training. In order to test how well noninvasive EMG signals can predict the biomechanics of human balancing movements, neural nets were trained using data from the lateral swaying tests. Training sets included data from both regular and random swaying tasks, whereas the neural nets were tested using only data from random tasks. The neural nets used were all feed-forward with two node layers. The first (input) layer had 50 nodes using a tangent sigmoid operator, and the second (output) layer had a single node using a linear operator. Each input data point for these neural nets was a 16 element vector giving EMG values for the 16 muscles monitored at a particular time increment. For this investigation, output data were the center of pressure (CP) measured by two force plates. Here output data were shifted in time to improve predictive capabilities. For example, using an input vector of EMG values corresponding to time increment i, an output (one-element) vector of the y component of CP, shifted ahead by 3 time increments (time increment i + 3), caused the neural net to be trained to predict the COP value 100 ms into the future (since the data increment corresponded to 30 Hz).

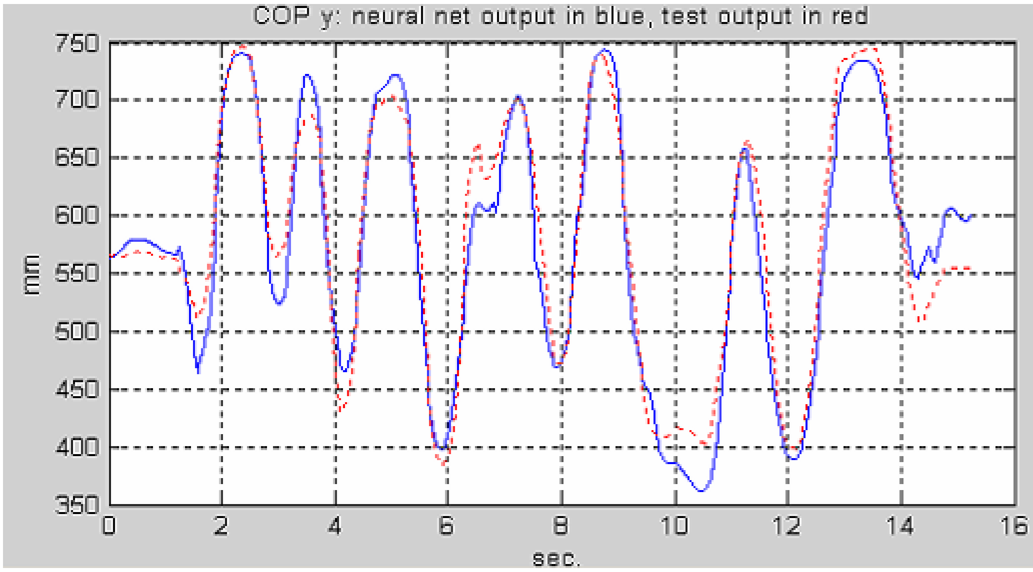


FIGURE 18: Center of pressure trajectory in the medial-lateral direction for simple swaying motions while in standing double support. An EMG-trained neural net (dotted line) makes an independent prediction of the center-of-pressure trajectory measured directly from a force plate (solid line). The model tracks the measured center-of-pressure data with an accuracy greater than 90%.

Results. A number of experiments were performed with shifts of 0, 33, 67, 100, and 133 ms into the future. The most accurate results were obtained with a 100 ms shift, indicating that the EMG signal leads the output signal by approximately this amount. Figure 18 demonstrates the predictive capability of a neural net trained to predict CP position 100 ms into the future based on EMG data. The average error is 2 cm, which is small given the overall range is 36 cm.

Discussion. These results indicate that EMG signals might be used in future O&P devices to anticipate the movement intent of a human user, making it possible to construct fully actuated O&P devices that respond faster than biological limbs. Perhaps future O&P control systems that employ distributed sensing architectures will enable life-like artificial appendages to respond seamlessly to the intent of the human user, increasing the controllability of such devices and the intimacy between human and rehabilitative mechanism.

5.4 Concluding Remarks

We are at the threshold of a new age when O&P appendages will no longer be separate, life-less mechanisms, but will instead be intimate extensions of the human body, structurally, neurologically and dynamically. Such a merging of body and machine will not only increase the acceptance of the physically-challenged into society, but will also enable individuals suffering from limb dysfunction to more readily accept their new artificial appendages as part of their own body, rather than foreign objects that must simply be tolerated. Several scientific and technological advances will accelerate this mergence. Critical will be the development of biomimetic humanoid control schemes that will enable the physically challenged to move with an enhanced level of biological realism (Schaal 1999, Kuo 1999, Hofmann et al. 2002), and the advancement of artificial or actin-myosin based actuators that can generate force quietly at a comparable efficiency, plasticity and contractility to *in vivo* skeletal muscle (Dennis & Herr 2000, Bar-Cohen et al. 2001). Still further, the improvement of internal prostheses that can safely measure perpheral and central neural signals within the body will also prove critical (Loeb 2001).

Key advancements in these research areas will most certainly have a dramatic effect on the quality of life of the physically challenged, but even dramatic research progress will not entirely duplicate the functionality of an intact biological limb. Many new technological horizons will emerge as engineers and scientists continue to make progress in areas relavant to the O&P field. One interesting area of future research will be to neurally communicate a sense of touch back to the user. It has been argued in this chapter that to assess the intent of the human user is a major technological problem that must be solved if artificial appendages are to respond seamlessly to the user's movement desires. An even greater challege in O&P development will be to comunicate the state of the artificial appendage back to the human as an afferant signal. In this case not only will the appendage respond to the intent of the user but the appendage itself will communicate exteroceptive and proprioceptive information back to the human, enabling a person that has suffered leg amputation to not only walk across a sandy beach, but to feel the sand against his cyborg feet.

5.5 References

Amis, A. A. (1990). Biomechanics of the Upper Limb: Forearm, Wrist and Fingers, Current Orthopaedics, 4, 107-111.

AMTI OR6-5 Biomechanics Platforms, http://www.amtiweb.com

Banerjee, N.S. (1982) Rehabilitation Management of Amputees. Williams and Wilkins, Baltimore, 99-149.

Bar-cohen, Y; Sherrit, S and Lih, S-S. (2001). Characterization of the Electomechanical Properties of EAP materials, Smart Structures and Materials 2001: Electroactive Polymers Actuators and Devices, Proceedings of the SPIE, 4329, 319-327.

Baughman, R.H., Shacklette, L.W., Elsenbaumer, R.L., Plichta, E. and Becht, C. (1990) Conducting Polymer Electromechanical Actuators, 559-582 -in- BREDAS, J.L. and CHANCE, R.R. (eds) (1990) Conjugated Polymeric Materials: Opportunities in Electronics, Optoelectronics, and Molecular Electronics, Kluwer Academic Publishers, Netherlands

Baughman, R.H. (1996) Conducting Polymer Artificial Muscles, Synthetic Metals, 78, 339-353

Bennet Wilson, A. Jr. (1989) Limb Prosthetics 6th Edition. Demos, New York, 27.

Blaya J. (2002). Force controllable ankle foot orthosis to assist drop-foot gait. Mech. Eng. MS Thesis, MIT.

Buckley, M.A., Yardley, A., Johnson, G.R. and Carus, D.A. (1996) Dynamics of thee Upper-Limb During Performance of the Tasks of Everyday Living – A Review of the Current Knowledge Base, Proceedings of the Institute of Mechanical Engineers, 210, 241-247.

Burger, H. and Marincek, C. (1994) Upper Limb Prosthetic Use in Slovenia, Prosthetics and Orthotics International, 18, 25-33.

Chao, E.Y. and Morrey, B.F. (1978) Three-Dimensional Rotation of the Elbow, Journal of Biomechanics, 11, 57-73.

Chao, Y.S., Kai-nan, A., Cooney, W.P. and Linscheid, R.L. (1989) The Biomechanics of the Hand: A Basic Research Study. World Scientific, London. 163-178.

Cone, R.O., Szabo, R., Resnick, D., Gelberman, R., Taleisnik, J. and Giluls, L.A. (1983) Computer Tomography of the Normal Radioulnar Joints, Investigative Radiology, 8, (6), 541-545.

Croney, J. (1980) Anthropometry for Designers, Batsford, London.

Datta, D. and ibbotson, V. (1998) Powered Prosthetic Hands in the Very Young, Prosthetics and Orthotics International, 22, 150-154.

Deffenbaugh B., Herr H., Pratt G., Wittig M. (2001). Electronically Controlled Prosthetic Knee. U.S. Patent Pending.

Della Santa, A., De Rossi, D. and Mazzoldi, A. (1997) Characterization and Modelling of a Conducting Polymer Muscle-Like Linear Actuator, Journal of Smart Materials and Structures, 6, 23-34.

Dennis R.G. Bipolar implantable stimulator for long-term denervated muscle experiments. *Med & Biol Eng & Comput Med & Biol Eng & Compu*t, March, 36: 225-28, 1998.

Dennis B., Herr H. (2001). An Actin-Myosin Machine. MIT Artificial Intelligence Laboratory. Research Abstracts. Pp. 241-242.

Derossi, D., Suzuki, M., Osada, Y. and Morasso, P. (1992) Pseudomuscular Gel Actuators for Advanced Robotics, Journal of Intelligent Materials, Systems and Structures, 3, (January), 75-95.

Dietl H., Bargehr H. (1997). Der Einsatz von Elektronik bei Prothesen zur Versorgung der unteren Extremitat. *Med. Orth. Tech.* 117: 31-35.

Gard, A. (1999). The influence of stance-phase knee flexion on the vertical displacement of the trunk during normal walking. *Archives of Physical Medicine and Rehabilitation.* Vol. 80.

Guyot, J. (1990) Atlas of Human Limb Joints Second Edition, Springer-Verlag, London, 164.

Hansen, A. H. (2002) Roll-over characteristics of human walking: With applications for artificial limbs. Ph.D. Dissertation, Northwestern University.

Hansen, A.H., Childress, D.S., and Knox, E.H. (2002) Prosthetic foot roll-over shapes with implications for alignment of trans-tibial prostheses. *Prosthetics & Orthotics International*, 24(3), 205-215.

Harris A., Miledi R. (1972) A study of frog muscle maintained in organ culture. *J. Physiol.*, 221: 207-226.

Herr H., Wilkenfeld A., Olaf B. (2001). Speed-Adaptive and Patient-Adaptive Prosthetic Knee. U.S. Patent Pending.

Hofmann A., Herr H. (2002). Predicting human biomechanics with EMG. MIT AI Lab Abstracts.

Hofmann A., Popovic M., Herr H. (2002). Humaniod Standing Control: Learning from Human Demonstration. *Journal of Automatic Control*. University of Belgrade. 12: 16-22.

Hunter, Ian W., Hollerbach, John M., and Ballantyne, John, (1991). A Comparative Analysis of Actuator Technologies for Robotics. Robotics Review 2, MIT Press.

Ibbotson, V. (1999) Private Communication with Ms. V. Ibbotson Senior Occupation Therapist at the Mobility and Specialised Rehabilitation Centre, Northern General Hospital, Sheffield.

Jacobsen SC, Knutti D.F., Johnson R.T., Sears H. H. (1982) Development of the Utah artificial arm. *IEEE Trans Biomed Eng* BME-29:249-269.

James K., Stein R.B., Rolf R., Tepavac D. (1990). Active suspension above-knee prosthesis. *Goh JC 6th Int Conf Biomech Eng* pp 317-320.

Jung, K.; Ryew, S.; Jeon, J. W.; Kim, H. and Choi, H. (2001) Experimental Investigations on Behavior of IPMC Polymer Actuator and Artificial Musclelike Linear Actuator, Smart Structures and Materials 2001: Electroactive Polymers Actuators and Devices, Proceedings of the SPIE, 4329, 449-457.

Kaneto, K., Kaneto, M., Min, Y. and Macdiarmid, A.G. (1995) Artificial Muscle - Electromechanical Actuators Using Polyaniline Films, Synthetic Metals, 71, 2211-2212.

Kapandji, I.A. (1982) The Physiology of the Joints Volume 1 The Upper-Limb. Churchill Livingstone, London.KAPIT, W. and ELSON, L.M. (1993) The Anatomy Colouring Book 2nd Edition. Harper Collins Publishers, New York.

Kastner J., Nimmervoll R., Kristen H., Wagner P. (1998). A comparative gait analysis of the C-Leg, the 3R45 and the 3R80 prosthetic knee joints. http://www.healthcare.ottobock.com.

Kinnier, A.B. (1965) Hendon Pneumatic Power Units and Controls for Prostheses and Splints, Journal of Bone and Joint Surgery, 47B, (3), 435-441.

Kostuik, J.P. (1980) Amputation Surgery and Rehabilitation (The Toronto Experience). Churchill Livingstone, New York.

Kuo, A. D. (1999). Stabilization of lateral motion in passive dynamic walking. *International Journal of Robotics Research.* 18(9): 917-930.

Kyberd, P. Algorithmic Control of a Multifunctional Hand Prosthesis. PhD Thesis, University of Southampton, 1990.

Landsmeer, J.F. (1976) Atlas of the Anatomy of the Human Hand. Churchill Livingstone.

Lawrence, A., De Rossi, D., Baughman, R. (1993) Application of Conducting Polymers in Medical Robotics and Prosthetics, IEEE Proceedings of the Annual Conference on Engineering in Medicine and Biology, 15, (2), 956-957.

Leonard, J.A. Jr.; Esquenazi, A.; fisher, S.V.; Hicks, J.E.; Meier, R.H. 3rd and Nelson, V.S. (1989) Prosthetics, Orthotics and Assistive Device. 1. General Concepts, Arch. Phys. Med. Rehabilitation, 70(5-S), S195-201.

Loeb, G. E. (2001). Neural prosthetics. The Handbook of Brain Theory and Neural Networks. M.A. Arbib (ed), MIT Press, Cambridge, Mass, 2nd ed.

Martin, R. (2000) Private Communication R. Martin Prosthetist at The Northern General Hospital Centre of Mobility and Specialised Rehabilitation, Sheffield.

McDonagh M. (1984) Mechanical Properties of muscles from xenopus borealis following maintenance in organ culture. *Biochem. Physiol.,* 77A: 377-382.

Nakamura, T. *et al* (1994) A Biomechanical Analysis of Pronation-Supination of the Forearm Using Magnetic Resonance Imaging: Dynamic Changes of the Forearm During Pronation-Supination, Journal of the Japanese Orthopaedic Association, 68, 14-25.

Norkin, C.C. and Levangie, L. (1992) Joint Structure and Function: A Comprehensive Analysis 2nd Edition. F.A. Davis Company, Philadephia.

Otero, T.F. and Sansinena, J.M. (1997) Bilayer Dimension and Movement in Artificial Muscles, Bioelectrochemistry and Bioenergetics, 42, 117-122.

Palmer M. (2002). Controlled plantarflexion: the spring-like response of the human ankle during unimpaired walking. Mech. Eng. MS Thesis, MIT.

Popovic M., Herr H. (2002). Conservation of Angular Momentum in Human Movement. MIT AI Lab Abstracts.

Popovic D., Schwirtlich L. (1988). Belgrade active A/K prosthesis. De Vries J (ed.) Electrophysiological Kinesiology. Excerpta Medica, Amsterdam, Int Congress Series No 804. pp 337-343.

Popovic D., and Sinkjaer T. (2000). Control of movement for the physically disabled. Springer-Verlag London.

Popović D., Tomović R., Tepavac D., Schwirtlich L. (1991). Control aspects in active A/K prosthesis. *Int J Man-Machine Studies* 35:751-767.

Pratt G. (2000). Legged Robots: What's New Since Raibert. *IEEE Robotics and Automation Magazine. Research Perspectives*. pp. 15-19.

Pratt, G., Williamson, M. (1995). Series Elastic Actuators. *Proceedings of IROS '95*, Pittsburgh, PA.

Rahman, T.; Sample, W.; Seliktar, R.; Alexander, M. and Scavina, M. (2000) A Body-Powered Functional Upper-Limb Orthosis, Journal of Rehabilitation Research and Development, 37(6), 675-680.

Reichardt, O (1978) Robots Fact Fiction and Prediction. Thames & Hudson, London.SAUTER, W.F. (1991) The Use of Electric Elbows in the Rehabilitation of Children With Upper-Limb Deficiences, Prosthetics and Orthotics International, 15, 93- 95.

Schaal S. (1999). Is imitation learning the route to humanoid robots? *Trends in Cognitive Sciences* 3:233-242.

Scott, R.N. and Parker, P.A. (1988) Myoelectric Prostheses: State of the Art, Journal of Medical Engineering and Technology, 12, (4), 143-151.

Skahen, J.R. et al (1997) The Interosseous Membrane of the Forearm: Anatomy and Function, The Journal of Hand Surgery, 22A, (6), 981-985.

Sheridan, T.B. and Mann, R.W. (1978) The Choice of Control System for People with Severe Motor Impairment, Human Factors, 20, (3), 321-338.

Smela, E., Inganas, O. and Lundstrom, I. (1995) Controlled Folding of Micrometer-Size Structures, Science, 268, (23 June), 1735-1740.

Smith, D.G. and Burgess, E.M. (2001) The use of CAD/CAM Technology in Prosthetics and Orthotics--Current Clinical Models and a View to the Future, Journal of Rehabilitation Research and Development, 38(3), 327-334.

Smith, K.L., Lawrence-weiss, E. and Lehmkuhl, D. (1996) Brunnstrom's Clinical Kinesiology, 5th Edition, F. A. Davis Company, Philidelphia.

Tomović R., Boni G. (1962). An adaptive artificial hand. *IRE Trans Autom Contr* AC-7:3-10.

Vicon Motion Systems, http://www.vicon.com

Weir R., Heckathorne C., Childress D. (2001) Cineplasty as a control input for externally powered prosthetic components. *Journal of Rehabilitation Research & Development*, Vol. 38 No. 4, 357-363.

Whiteley, G.P. An Articulated Skeletal Analogy of the Human Upper-Limb, PhD Thesis, Sheffield Hallam University, 2000.

Wilkenfeld A. (2000). An Auto-Adaptive External Knee Prosthesis. PhD Thesis, MIT.

Wilkie, D.R. (1960). Thermodynamics and the interpretation of biological heat measurements. In: Progress in Biophysics and Biophysical Chemistry Vol 10., Ed. J.A.V. Butler and B. Katz. MacMillan Co., NY. 10, 260-298.

Woledge R., Curtin N., Homsher E. (1985). Energetic Aspects of Muscle Contraction. Monographs of the Physiological Society No. 41. Academic Press.

Youm, Y., Dryer, R.F., Thambyrajah, K., Flatt, A.E. and Sprague, B.L. (1979) Biomechanical Analyses of Forearm Pronation-Supination and Elbow Flexion-Extension, Journal of Biomechanics, 12, 245-255.