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Characterization of Ankle Function during Stair Ambulation

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Abstract— **The aim of this study was to examine the ankle joint during level walking, stair ascent, and stair descent to determine models for use in the design of prosthetic and orthotic systems. Ten healthy subjects were asked to walk (1) across a level walkway, (2) up, and (3) down an instrumented stairway. Sagittal plane kinematic and kinetic data were analyzed to obtain ankle biomechanics during the stance phase of each task. Each stance phase was broken down into sub-phases based on the power trajectory. The ideal model was taken to be the simplest combination of mechanical elements (springs, dampers, and torque actuators) that could reproduce the patterns observed in ankle biomechanics. Besides, we studied the transitions from level walking to stair ascent and from stair descent to level walking and showed that mechanical elements can be used to model these transitions as well. These results are promising to the design of next generation ankle orthotic and prosthetic systems because they show that relatively simple mechanical elements can be utilized to mimic ankle biomechanics.**

Keywords- **Ankle biomechanics, orthotics, prosthetics, stair ambulation**

I. Introduction

Currently available ankle orthotic and prosthetic (O&P) systems are passive devices whose characteristics are set to maximize mobility during level walking. Throughout the course of a day, O&P users perform many more ambulatory tasks, including walking up and down stairs, walking up and down ramps, and walking on uneven surfaces. In order to maximize user's mobility, O&P systems should adapt to the tasks in which the user engages. While systems should eventually adapt to any ambulatory task, the focus of this research is on modeling ankle biomechanics during stair walking, as stairs are frequently encountered in daily life.

The literature is replete with examples of how characterizing a biomechanical system by means of a model can be a tool for understanding the function of that system in a given situation [1-9]. Modeling ankle biomechanics using simple mechanical devices is appealing to those attempting to design ankle-foot O&P devices that mimic biology. Palmer previously performed a similar study for level walking at different gait speeds [10]. He modeled the stance phase of the gait cycle by separating it into three distinct sub-phases and using simple mechanical elements, namely springs, dampers, and torque actuators. These results were then used at MIT to design an active ankle-foot orthosis with variable impedance for the treatment of drop foot [11]. Results showed that the orthosis was able to significantly reduce the occurrence of slap-foot. This research shows that promising results can be obtained by relying on modeling and active control.

As walking plays an integral role in our daily lives, it has been extensively researched and most of this research has been focused on level walking performed in a 'perfectly flat laboratory' [12]. As biomechanics research focuses around level walking, so do the models developed. This work attempts to expand upon the models developed by Palmer to include stair walking and the transitions between level walking and stair ambulation.

II. Methodology

A. Subjects

Ten healthy young adults (23-29 years, mean 25 years of age; 52.3-73.4 kg, mean 63 kg in mass; and 160-182.5 cm, mean 170.6 cm in height) participated in this research. Subjects had no neurological, musculoskeletal or chronic ankle or knee problems. Prior to participation in the study, written informed consent was obtained from each subject.

B. Equipment

Kinematic and kinetic data was obtained using an 8 camera Vicon motion capture system (VICON 512, Oxford Metrics, Oxford, UK) and two AMTI force platforms. Data was processed using Vicon commercialized software (Vicon Plug-in-Gait). A customized three-step stair system built by AMTI (Watertown, MA) was used for this study. The stairs are 7 inches high, 11.5 inches deep and 3 ft wide, in accordance with Massachusetts state building codes. The forces on the steps were measured using the two AMTI force platforms (see Fig 1). The lowest step is attached to the second platform, the middle step to the first platform and the highest step to the second platform.

C. Procedure

Testing was performed at the Motion Analysis Laboratory at Spaulding Rehabilitation Hospital. 3 D pelvic and bilateral lower extremity joint kinematics and kinetics were collected during (1) level walking, (2) stair ascent, and (3) stair descent. The camera system measured the three-dimensional position, at 120 frames per second, of reflective markers attached to lower extremity bony landmarks. Ground reaction forces were measured synchronously with

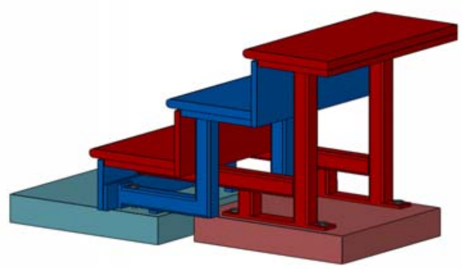

Fig. 1 Schematic of custom-built stair system.

the kinematic data using two staggered force platforms. Ten repetitions including complete motion analysis data were collected and average lower extremity biomechanical values for each subject were obtained. Joint kinetics in each plane was calculated using a lower body inverse dynamic model (Vicon BodyBuilder, Oxford Metrics) and data was normalized for body weight. Kinetic data during stair walking was calculated using a modified Vicon model to account for the stair height. Temporal parameters were obtained using the force platform and kinematic information to define foot contact times and distance parameters.

D. Modeling

The data derived using a modified BodyBuilder model, including ankle angular position, and ankle torque, were then analyzed using a custom program written in MATLAB (The MathWorks, Natick, MA). Ankle angular velocity time histories were found by numerically differentiating the relevant angle components as determined by the Plug-in-Gait software. Ankle powers were estimated by multiplying the normalized ankle moments by the angular velocities of the ankle joint. The net energy gained or lost in the ankle during stance was estimated by determining the area under the ankle moment vs. ankle angle curve using a trapezoidal approximation [13].

Employing the same approach as Palmer [10], ankle function was characterized by considering ankle angular position and velocity as inputs into a 'black box' and ankle torque as the output. Kinematic and kinetic data from 10 healthy subjects were used to find the simplest combination of mechanical elements to produce the observed position/torque and velocity/torque relationships. These mechanical elements represent the sum effect of the lengthening and shortening of the muscles, tendons, and ligaments as well as deformation of the foot and any other mechanisms of generating torque about the ankle joint [10].

In order to supplement the model of level walking described by Palmer, the same set of mechanical elements were considered for stair walking. These mechanical elements include torsional springs, torsional dampers, and torque actuators. Of these elements, springs and dampers are passive elements, while torque actuators are the only active elements. Torque actuators were only considered for periods where the power at the ankle was positive and the amount of positive work done was greater than any energy that might have been stored by the ankle system [10].

Stance during each activity was divided into periods based on whether power was generated or absorbed. Since it would be more cumbersome to design a device that changed behavior as a function of power, changes in the position and velocity were used to identify the beginning and end of each phase.

During periods where the ankle was characterized by spring-like behavior, the relationship between ankle torque and ankle angular position was expected to satisfy the equation

$$\tau = k \cdot \theta + \tau_0 \tag{2.1}$$

where τ is the ankle torque, k is the stiffness, θ is the ankle angular position, and τ_o is the ankle torque when the position is zero. Values for these parameters were determined using linear regression and the adequacy of the model was assessed by the coefficient of determination (r^2) and viewing scatter plots of the residuals vs. angular position. If the coefficient of determination was low, or the residuals were highly patterned, the linear model was deemed insufficient [10].

During periods where the ankle was characterized by damper-like behavior, the relationship between ankle torque and ankle angular velocity was expected to satisfy the equation

$$\tau = b \cdot \left(\frac{d\theta}{dt}\right) + \tau_0 \tag{2.2}$$

where τ is the ankle torque, b is the damping constant, $d\theta/dt$ is the ankle angular velocity, and τ_o is the ankle torque when angular velocity is zero. Values for these parameters were found by performing a linear regression of the ankle torque vs. ankle velocity. The adequacy of the model was determined in the same way as for spring-like behavior. For both springs and dampers, when the fitting wa not satisfactory, we considered the use of non-linear elements.

III. RESULTS

A. Level Walking

Characterization of the ankle during level walking was done in three sub-phases: (1) controlled plantarflexion (CP), (2) controlled dorsiflexion (CD), and (3) powered plantarflexion (PP), as described by Palmer [10]. Our results were consistent with Palmer's work as CP was modeled as a linear spring, CD as a hardening, non-linear spring, and PP as a torque actuator (a torque actuator was necessary because more energy was needed than stored by the spring during CD).

B. Stair Ascent

Because ankle power changes sign three times during stance, stair ascent was separated into four subphases. These phases consisted of two controlled dorsiflexion phases, where power was absorbed, and two powered plantarflexion phases, where power was generated. Phases were defined by zero crossings of ankle velocity, as shown in Figure 2. Since power is defined as the product of moment (i.e., torque) and velocity, and moment is always negative, these points coincided with the zero crossings of the power trajectory.

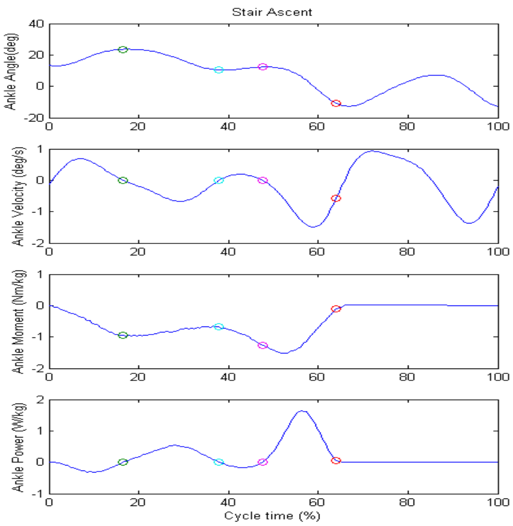

Fig. 2 Ankle angle, angular velocity, moment and power for one cycle of stair ascent from a representative trial. Circles indicate the zero crossings of the velocity that divide the stance phase into controlled dorsiflexion 1, powered plantarflexion 1, controlled dorsiflexion 2, and powered plantarflexion 2.

The first subphase of stair ascent is controlled dorsiflexion 1 (CD1). During this phase, the power is always negative, thus only passive devices where considered. Linear regression of the ankle torque on the ankle angular position during CD1 was significant for all trials. This indicates that during this phase, the ankle behaves as a linear spring. Numerical analysis showed that the stiffness is approximately 0.1 N m / kg deg.

The next subphase is powered plantarflexion 1 (PP1). Since less than half of the work required for the PP1 phase is absorbed during the CD1 phase, it was not possible to use a purely passive device to drive this phase. Using the same spring from the CD1 phase could move the ankle through the beginning of the PP1 phase, but an active torque actuator was needed to assist the spring.

The third subphase is controlled dorsiflexion 2 (CD2). During this phase, the power is always negative so only passive devices were considered. Scatter plots of ankle moment vs. ankle angle showed a highly linear relationship between the two variables, characterized by coefficients of determination (r^2) greater than 0.90 for all subjects and no obvious patterns in the residuals.

The final subphase is powered plantarflexion 2 (PP2). The amount of energy absorbed during the CD2 phase was, on average, only one tenth of the amount needed during the PP2 phase. Therefore, only active devices were considered. Scatter plots of ankle torque vs. ankle angular position showed a strong linear relationship between the two variables with the exception of an initial subphase lasting approximately 50 ms during which a nonlinear behavior was apparent. Therefore, this phase (i.e., PP2) was modeled using a torque actuator controlled by ankle angular position marked by a nonlinear transfer function for the initial 50 ms and by a linear transfer function for the rest of PP2.

C. Stair Descent

Stair descent was separated into three subphases: controlled dorsiflexion 1 (CD1), controlled dorsiflexion 2 (CD2), and powered plantarflexion (PP), as shown in Fig 3.

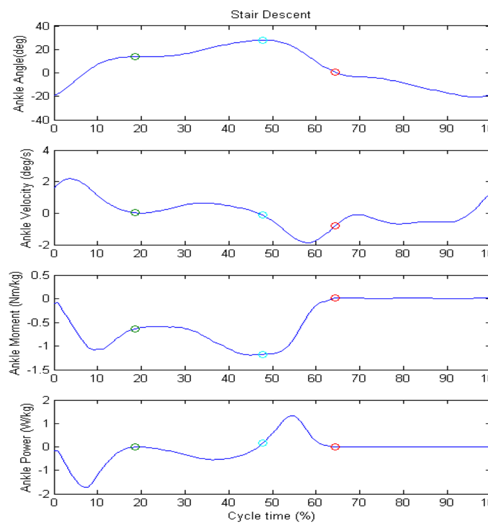

Fig. 3 Ankle angle, angular velocity, moment and power for one cycle of stair descent from a representative trial. Circles indicate the zero crossings of the velocity that divide the stance phase into controlled dorsiflexion 1, controlled dorsiflexion 2, and powered plantarflexion.

During CD1, power is absorbed and as such, only passive devices (dampers and springs) were considered when modeling this phase. Because CD1 is followed by another energy storage phase (CD2), springs were not considered an ideal model for CD1 since an excessive amount of energy would need to be dissipated during the following phases. Therefore, a damper was preferable. However, this phase could not adequately be modeled using a simple linear damper; therefore, a nonlinear damper was considered. It is worth emphasizing that nonlinear dampers can be implemented using magnetorheological fluids. These dampers can vary their behavior in any desired pattern by modulating the magnetic field applied to the fluid.

During the second phase, CD2, the power was negative for a majority of the phase. Therefore, only springs and dampers were considered for this phase. Since the energy absorbed in this phase is greater than the amount needed in the third phase, PP, it was most efficient to view these phases as the energy storage and release of a spring with additional damping. Scatter plots of ankle torque vs. ankle angle revealed a significant relationship between the two variables.

The final subphase is PP. To utilize the energy release from the CD2 phase, the PP phase was modeled as a spring. The scatter plot of ankle torque vs. ankle angle during the PP phase shows a relationship between the two and examination of the residuals showed a strong linear correlation with angular velocity. Therefore it was found that ankle behavior in this phase can be adequately modeled as a spring and a damper.

While these models are the best fit for each phase individually, the stiffnesses required in the CD2 and PP phases are not the same. In order to get an energy return, the same spring must be used for both phases. Since the stiffness required in the second phase is greater than that of the first, the entire system could be modeled using a linear spring (with a single stiffness value) and some dampening.

D. Transitions

In addition to modeling the stance phase of level walking, stair ascent, and stair descent, it is necessary to look at the transitions between tasks in order to adequately design orthotic and prosthetic (O&P) devices.

The transition from level walking to stair ascent was observed during the stance phase of the foot contacting the plate (prior to the steps). During CP, ankle function was characterized by a linear spring as it was in level walking. During CD, the moment exhibited a biphasic pattern, unlike walking, and significantly less power was absorbed than in level walking. During the final phase, PP, the ankle behaves in the same way it does during stair ascent. The energy to supply this phase must be generated by a torque actuator. During this time, the ankle torque is directly proportional to the ankle position.

The transition from stair descent to level walking was observed as the foot coming from the second step struck the force platform on the ground in front of the steps. Since power changed sign three times, the transition was split into three phases. The first phase, controlled dorsiflexion, was identical to the CD1 phase of stair descent. During the second phase, mid-stance, the power was predominately positive and was therefore modeled by a torque actuator. During the final two phases, ankle behavior was identical to that of level walking.

V. CONCLUSION

In this work, we showed that ankle biomechanics during stair ambulation can be modeled using simple mechanical elements. We showed that ankle function can be separated into phases using information from simple devices (i.e., sensing ankle angle and velocity). Each of these phases can be modeled using mechanical elements. Wherever simple linear elements (springs, dampers, and torque actuators) were insufficient, the ankle was modeled using nonlinear elements, which can be implemented as well using existing technology. This is significant because it shows that it is possible to create intelligent, next generation, O&P devices that can alter their properties in accordance to the ambulatory task being performed by the user. However, in order to make this usable in an actual O&P device, there has to be a way to switch from one mode to another. We showed herein that the transitions from level walking to stair ascent and from stair descent to level walking can be modeled to provide a smooth transition from one ambulatory task to the next. In doing so, it is possible to achieve intelligent O&P devices that not only function in distinct states according to the task the user is engaged in, but also can transition between states smoothly and effectively.

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