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Estimation of Ground Reaction Force and Zero Moment Point on a Powered Ankle-Foot Prosthesis

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Abstract— **The ground reaction force (GRF) and the zero moment point (ZMP) are important parameters for the advancement of biomimetic control of robotic lower-limb prosthetic devices. In this document a method to estimate GRF and ZMP on a motorized ankle-foot prosthesis (MIT Powered Ankle-Foot Prosthesis) is presented. The method proposed is based on the analysis of data collected from a sensory system embedded in the prosthetic device using a custom designed wearable computing unit. In order to evaluate the performance of the estimation methods described, standing and walking clinical studies were conducted on a transtibial amputee. The results were statistically compared to standard analysis methodologies employed in a gait laboratory. The average RMS error and correlation factor were calculated for all experimental sessions. By using a static analysis procedure, the estimation of the vertical component of GRF had an averaged correlation coefficient higher than 0.94. The estimated ZMP location had a distance error of less than 1 cm, equal to 4% of the anterior-posterior foot length or 12% of the medio-lateral foot width.**

I. INTRODUCTION

Transtibial amputees experience several clinical problems while wearing conventional passive prostheses. Some of these difficulties include deficient stability during standing and walking, asymmetric gait, slower walking speed, as well as increased metabolic rates when compared to intact persons [1]-[3]. To overcome some of the mentioned problems, researchers have proposed the use of motorized (active) prostheses which can actively control the ankle-foot joint and also provide mechanical power during walking [4]. One of the main obstacles in the development of these powered assistive devices is the design and implementation of prosthetic control strategies that can resemble the biological behavior of the human ankle-foot complex. A suggested approach to obtain such biomimetic control strategies is the incorporation of analytic methods and sensing technologies used in human biomechanic studies as well as in humanoid robotics.

In the fields of human biomechanics and humanoid robotics, postural control is critical for understanding balance and locomotion. Powerful control strategies for bipedal systems, proposed in both fields, rely on the knowledge of the ground reaction force (GRF) (i.e. force of interaction between the foot and the ground) and the zero moment point (ZMP) [5]. In the study of bipedal locomotion on level ground, the ZMP is defined as the ground reference point that corresponds to the center of pressure (COP)[10].

In biomechanics, several control methods that emphasize the ankle-foot joint behavior to understand the dynamics of human balance have been proposed [6]-[10]. In the area of humanoid machine control, ankle-foot joint control methods have been implemented to maintain the stability of legged robotic systems [8][11]-[16]. For these control schemes, ankle-foot movements are actively controlled to reposition the ZMP beneath the foot. By modulating the ZMP location, the ground reaction force can effectively be controlled, hence obtaining dynamic balance stability [10][17].

In order to estimate these parameters (GRF and ZMP) different technologies are used in both disciplines. In legged robotics the GRF and ZMP are generally measured with a series of sensors embedded in the robot's feet [13][14][31][32]. In human biomechanics, standard measuring techniques for the GRF and ZMP are restricted to a laboratory setting, where analysis tools, such as video motion capture and calibrated force platform systems, are available. However, several researchers have investigated how to develop wearable gait analysis tools that accurately estimate the GRF components and ZMP [18]-[23]. Moreover, sensing technologies have also been implemented in prosthetic devices for gait studies and prosthetic assessment [26]-[30].

The goal of present research is to implement and evaluate a method to estimate the GRF and ZMP employing an autonomous powered ankle-foot prosthesis. In addition, the presented autonomous system can be used as gait assessment tool to better understand amputee ambulation in real environments outside a laboratory. With this project we hope to further contribute to the development of biologically realistic lower-limb assistive devices that improve amputee locomotion.

First, in this paper, the instrumentation of the MIT powered ankle-foot prosthesis is presented. Then, the experimental and analytical methods to estimate the GRF and ZMP are described. Finally, the estimation results compared to the values obtained with standard gait analysis methods is discussed.

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II. SYSTEM DESIGN AND IMPLEMENTATION

A. Powered Ankle-Foot Prototype

A novel motorized ankle-foot prosthesis prototype [4], was instrumented in order to determine the ZMP and GRF in transtibial amputee gait. This prosthesis is based on a force controllable actuator, called Series-Elastic Actuator (SEA), originally developed for legged robots [33].The purpose of this biomimetic prosthetic device is to emulate the behavior of the normal human ankle-foot complex. It can provide mechanical power and varying stiffness depending on walking speed and gait phase of the amputee. With this technology it is possible to provide a more natural gait and normal levels of energy expenditure of below knee amputees [35].

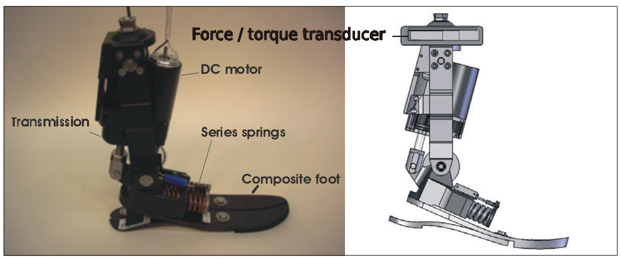

Figure 1. Powered Ankle-Foot Prosthesis

B. Instrumentation

The powered ankle-foot prosthesis is instrumented with a six-directional force - torque transducer from ATI Automation® (model Delta calibration SI660-60) installed at the shank level, 12 cm proximal to ankle joint. This transducer is connected through an interface power supply box to a host desktop PC computer with a 16 bit data acquisition card (National Instruments PCI 6220). The force information is recorded with this device at a sampling rate of 1000 Hz.

The ankle angle is measured by 500 counts/turn quadrature encoder module (US digital, Inc. HEDS 9140) mounted at the ankle joint. The torque around the prosthetic joint is measured indirectly with a linear potentiometer of 5 KΩ (Bourns 3048), installed across springs that are in series with the DC motor. As force is exerted from this series elastic actuator configuration [34] the ankle joint series springs suffer a linear compression, which measured by the variable resistor, provides joint- torque information. The signal from the linear potentiometer is conditioned using an analog low pass filter with a cut off frequency of 1.5 KHz. To determine foot-floor contact, a series of six force sensing resistors ® (Interlink Electronics No. 402) were embedded underneath the carbon composite foot of the prosthesis. Two sensors were mounted underneath the toe area, two in the metatarsal zone and two more on the heel of the leaf spring structures. An un-tethered wearable computing unit controls the powered prosthesis based on the monitored sensory information provided by the incorporated sensors.

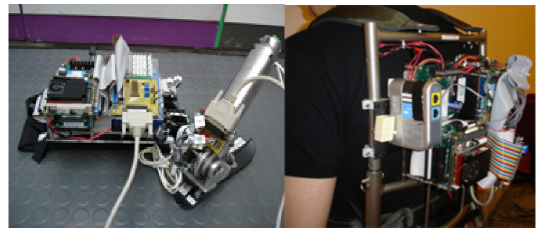

Figure 2. Wearable computing unit

The onboard computer of the wearable system is a MICROSPACE PC104 Pentium III CPU at 700 MHz (MSMP3XEG) from Advanced Digital Logic, Inc.). A multifunctional Input / Output module on PC104 format, from Sensory Co. Inc. (Model526) receives and processes all sensory information. The full system runs the Matlab® Kernel for xPC target applications. The actuator is driven by a motor amplifier (Accelnet Panel ACP-090-36) from Copley Controls Corp. The onboard computer runs the real-time software and data acquisition routines to control the powered-ankle.

III. EXPERIMENTAL METHODS

Standing and walking clinical studies were conducted on a transtibial amputee to evaluate the performance of the estimation methods. During the studies, kinetic and kinematic data were recorded with an AMTI© force platform and a 16 camera motion capture system VICON 810i (Oxford Metrics ®, Oxford, UK) respectively. Sampling rate of force plate information was 960 Hz and for motion data was 120 Hz. For the motion capture, nine reflective markers were mounted on ankle-foot prosthesis: Two markers on the heel region, three markers in the forefront of the foot and four markers on the force torque transducer. Simultaneous recording of sensory information was done by three computers. A *Control PC* recorded the active-ankle sensory information. A *Monitor PC* recorded independently force/torque sensor independently and a *Validation PC* recorded the kinematic and kinetic data of each experiment. For the standing studies, the participant was asked to perform twenty single leg standing trials (maintaining a relatively rigid posture) and twenty trials of anterior-posterior sway on a single leg support. Twenty more slow-walking trials were conducted.

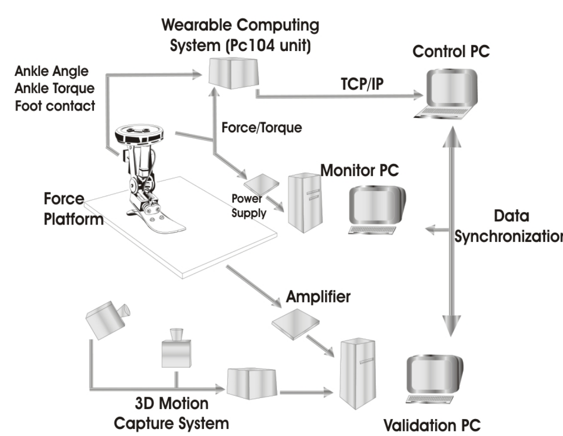

Figure 3. Schematic arrangement of the experimental set-up.

A. Static model analysis

A static model is proposed to estimate the ground reaction force vector and location of the zero moment point in the active ankle foot prosthesis. The estimation contemplates only single stance support. The estimated values are compared to the GRF, and COP provided by the force platform in conjunction with the kinematic data of the 3D motion capturing system.

The coordinate system employed for the model is Cartesian. The sign convention for the ground reaction force is based on [1][4], where the ground reaction force is positive upward and forward. The reference frames employed are:

a) ***{G}*** - **($O_GX_GY_GZ_G$)** - Global reference frame (motion capture frame (fixed).
b) ***{A}*** - ***($O_AX_AY_AZ_A$)*** - Force plate reference frame (fixed).
c) ***{S}*** - **($O_SX_SY_SZ_S$) -** Six axis force / torque sensor frame.
d) ***{P}*** - **($O_PX_PY_PZ_P$)** –Zero moment point /center of pressure reference frame (lies within the support polygon formed by the parts of the body in contact with the ground. i.e. prosthetic foot.

B. Estimation of Ground Reaction Force

In this analysis, the foot/ankle complex was considered a rigid body in static equilibrium. Forces measured by the force-torque sensor referenced at ***{S}*** can be mapped to the corresponding ZMP frame ***{P}*** (see figure 4). Assuming negligible inversion-eversion of the foot-ankle joint system:

$$\left[{}^{P}\vec{F}_P\right] = \left[{}^{P}_{S}\vec{R}\right]\left[{}^{S}\vec{F}_S\right]$$

${}^{P}_{S}\vec{R}$ is a rotation and translation matrices respectively, that relates frames ***{S}*** relative to ***{P}*** .

${}^{S}\vec{F}_S$ is the force vector at the load cell frame ***{S}*** and ${}^{P}\vec{F}_P$ is the mapped force vector at the ZMP frame ***{P}*** .

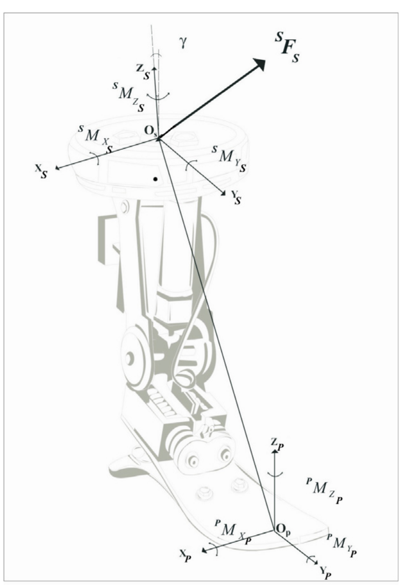

Figure 4. Ankle-foot complex free body diagram to estimate GRF

The forces that interact with the ground, relative to the ZMP frame ***{P},*** can then be expressed as:

$$ {}^{P}F_{X_S} = {}^{S}F_{X_S} \qquad (1)$$

$$ {}^{P}F_{Y_S} = {}^{S}F_{Y_S}\cos\gamma - {}^{S}F_{Z_S}\sin\gamma \qquad (2)$$

$$ {}^{P}F_{Z_S} = {}^{S}F_{Y_S}\sin\gamma + {}^{S}F_{Z_S}\cos\gamma \qquad (3)$$

Here the left superscript represents the reference frame and the right subscript represents the co-ordinate direction. The estimated forces represented relative to frame ***{P}*** should be represented relative to frame ***{A}*** (force plate reference frame) in order to compare and validate the results. The kinematic information to reference both systems is obtained using the 3D motion capture data.

The resulting force representation of the components of the GRF measured in the sensor frame ***{S}*** relative to the force plate frame ***{A}*** are:

$$ {}^{A}F_{X_A} = {}^{S}F_{X_S}\cos\alpha + {}^{S}F_{Y_S}\cos\gamma\sin\alpha - {}^{S}F_{Z_S}\sin\gamma\sin\alpha \qquad (4)$$

$$ {}^{A}F_{Y_A} = -{}^{S}F_{X_S}\sin\alpha + {}^{S}F_{Y_S}\cos\gamma\cos\alpha - {}^{S}F_{Z_S}\sin\gamma\cos\alpha \qquad (5)$$

$$ {}^{A}F_{Z_A} = {}^{S}F_{Y_S}\sin\gamma + {}^{S}F_{Z_S}\cos\gamma \qquad (6)$$

C. Estimation of Zero Moment Point

The area of the foot in contact with the ground surface is referred to as the ground support base (*g.s.b.)*. During single stance, the only external force acting on the g.s.b is assumed to be the GRF. The only area in contact with the ground is considered to be the g.s.b, which describes the support polygon. In figure 5) the free body diagram to determine the ZMP location is shown, where the foot/ankle is assumed to be a rigid body in static equilibrium.

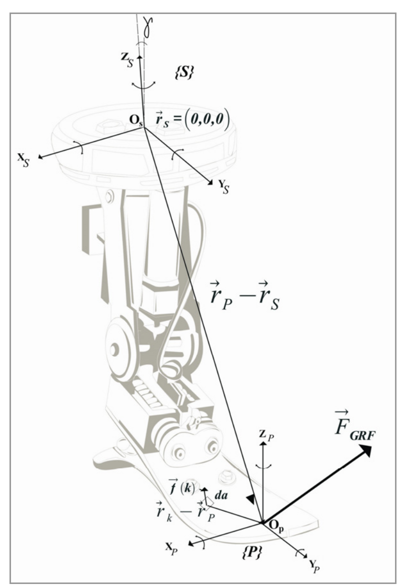

Figure 5. Ankle-foot complex free body diagram to estimate ZMP

At O_P the moment can be defined as:

$$\overrightarrow{M}_P = \int_{gsb} \left(\vec{r}_k - \vec{r}_P\right) \times \overrightarrow{f}(k)\, da \tag{7}$$

$\overrightarrow{M}_P$ = Moment around system ***{P}*** whose origin is O_P;
$\vec{r}_k$ is the location of a point k in the ground support base (g.s.b.) ;
$\vec{r}_P$ is the vector that defines the position of the zero moment point at O_P;
$\overrightarrow{f}(k)$ is the force acting on the point k ;
da is an infinitesimal element of the support surface.

At Os the moment can be defined as:

$$\overrightarrow{M}_S = \int_{gsb} \left(\vec{r}_k - \vec{r}_S\right) \times \overrightarrow{f}(k)\, da \tag{8}$$

$\vec{r}_S$ is the vector that defines the location of O_S and $\vec{r}_K - \vec{r}_S$ is the vector that goes from the point of application of the force $\overrightarrow{f}(k)$ to this point .

This moment can also be expressed as:

$$\begin{aligned}\overrightarrow{M}_S &= \int_{gsb} \left\{\left(\vec{r}_K - \vec{r}_P\right) + \left(\vec{r}_P - \vec{r}_S\right) \times \overrightarrow{f}(k)\right\} da \\ &= \overrightarrow{M}_P + \left(\vec{r}_P - \vec{r}_S\right) \times \int_{gsb} \overrightarrow{f}(k)\, da\end{aligned} \tag{9}$$

$\int_{gsb} \overrightarrow{f}(k)\, da = \overrightarrow{F}_{GRF}$ is the integration of all ground reaction force vector applied at the zero moment point
From the static equilibrium assumption the moment at frame ***{S}*** is:

$$\overrightarrow{M}_S = \overrightarrow{M}_P + \left(\vec{r}_p - \vec{r}_s\right) \times \overrightarrow{F}_{GRF} \tag{10}$$

If the origin for this analysis is at $\boldsymbol{O_S}$ then:

$$\overrightarrow{r_S} = \left[r_{SX_S}\ r_{SY_S}\ r_{SZ_S}\right] = [0\ 0\ 0] \tag{11}$$

By definition, at the ZMP, the torque around the horizontal axes is zero:

$$\overrightarrow{M}_P = \left[0\ 0\ {}^{P}M_{Z_S}\right] \tag{12}$$

Based on the components of $\overrightarrow{M}_S$ we then determine the vector $\overrightarrow{r_P} = \left[r_{PX_S}\ r_{PY_S}\ r_{PZ_S}\right]$ which is the location of the zero moment point in relation to the sensor reference frame ***{S}***:

$${}^{S}r_{ZMP\,X_S} = r_{PX_S} = \frac{-{}^{S}M_{Y_S} - \left(r_{PZ_S}\,{}^{S}F_{X_S}\right)}{{}^{S}F_{Z_S}} \tag{13}$$

$${}^{S}r_{ZMP\,Y_S} = r_{PY_S} = \frac{-{}^{S}M_{X_S} - \left(r_{PZ_S}\,{}^{S}F_{Y_S}\right)}{{}^{S}F_{Z_S}} \tag{14}$$

${}^{S}M_{Y_S}$, ${}^{S}F_{Y_S}$ - Moment and force, respectively around the $\boldsymbol{Y_S}$ axis of frame ***{S};***
${}^{S}M_{X_S}$, ${}^{S}F_{X_S}$ - Moment and force, respectively around the $\boldsymbol{X_S}$ axis of frame ***{S};***
${}^{S}F_{Z_S}$ - Force in the $\boldsymbol{Z_S}$ direction of frame ***{S};***
r_{PZ_S} - Distance in $\boldsymbol{Z_S}$ to the zero moment point. For the active system this value was calculated as be -$200\cos\gamma$ [mm]. This is assuming only rotation in the sagittal plane with no adduction - abduction in the joint.

IV. RESULTS

For each one of the experimental standing and walking trials the RMS error and the correlation coefficient (R) were calculated; these metrics compare the estimated parameters (using the static analysis described) with the validation measurements of GRF and ZMP location. The validation measurements were obtained from force plate and motion capture data. The GRF components are compared in the force plate reference frame ***{A}***. The vertical component of the force is FZ, the fore-aft component is FY and the medio-lateral component is FX. The ZMP is compared relative to the global reference frame ***{G}.***

Tables 1) and 2) summarize the obtained results comparing the GRF component and ZMP. These values are the mean and standard deviation of the RMS error as well as the average of the correlation coefficient of all standing and walking trials. Figures 6) and 7) show a comparison between estimated and measured ZMP trajectories during a representative experimental trial for standing with anterior-posterior sway. Figure 8) shows a comparison of the average vertical GRF component of the walking trials.

In summary, the estimation of the vertical component of GRF had a correlation coefficient higher than 0.94, averaging standing and walking trials. The estimated ZMP location had an average distance error of less than 1cm, equal to 4% of the anterior-posterior foot length or 12% of the medio-lateral foot width.

	RMS ERROR (N)		
Parameter	**MEAN**	**STDV**	**R**
FZ	1.5845	0.8619	0.8990
FY	3.2950	0.7586	0.3179
FX	7.2200	3.9862	0.3308

	RMS ERROR (mm)		
Parameter	MEAN	STDV	R
ZMP Y	2.1222	0.4272	0.9207
ZMP X	2.4732	0.2238	0.9121

Table 1. Average results for single leg standing trials. Active ankle-foot average distance between estimated and measured ZMP location: 2.911 mm

	RMS ERROR (N)		
Parameter	**MEAN**	**STDV**	**R**
FZ	3.5451	0.5454	0.9984
FY	4.4954	1.7912	0.5416
FX	13.399	2.8315	0.4671

Parameter	**RMS ERROR (mm)**		**R**
	MEAN	STDV	
ZMP Y	5.4388	1.3853	0.9504
ZMP X	9.0299	1.6693	0.9859

Table 2. Average results for slow walking trials (single support phase). Average distance between estimated and measured ZMP location: 9.01 mm

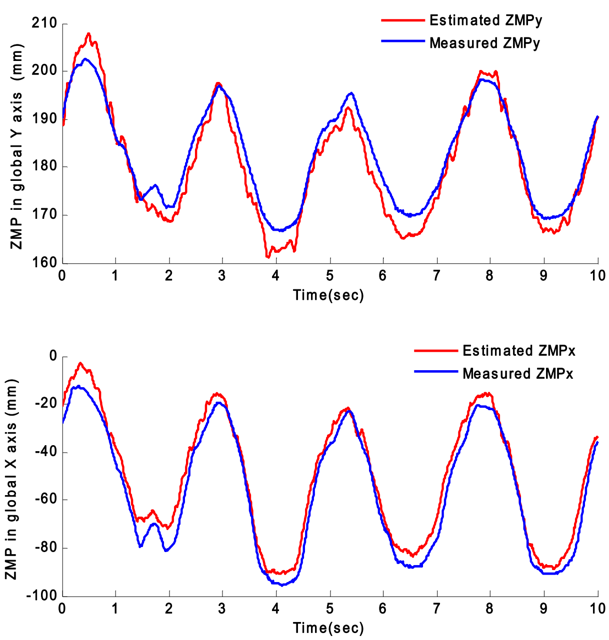

Figure 6. ZMP trajectory relative to the X and Y axis of the global coordinate frame {G}. Representative trial of single leg standing with anterior-posterior sway.

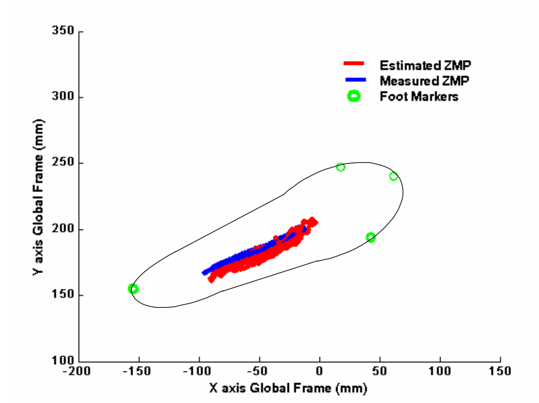

Figure 7. ZMP trajectory in global coordinate frame and its relation with the foot support polygon delimited by the four reflective markers.

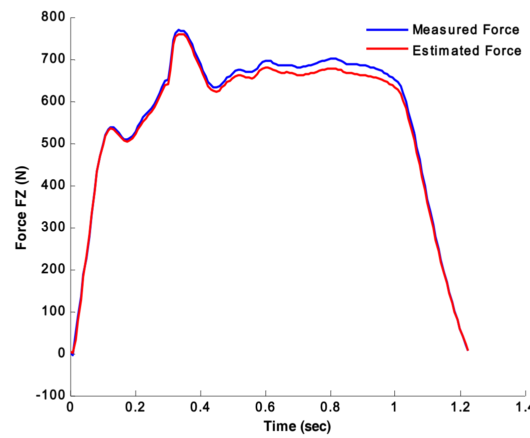

Figure 8. Average vertical GRF component FZ from heel strike to toe off during walking trials.

V. DISCUSSION

The observed results of the estimated vertical components of the GRF, when compared to the gait lab measured data, averaged a correlation factor higher than 0.9 across all experimental trials. In addition, the average RMS error was less than 5 N. The difference in the estimated and measured values of this component is primarily due to the static assumptions of the model. No dynamic behaviors were incorporated into the model. The shear (horizontal) forces had larger errors but had a small impact on the estimations of the ZMP during the standing and walking evaluations.

The spring-like behavior of the compliant foot prosthesis acts like a shock absorber for the forces interacting with the ground. This last can be a reason for discrepancy between measured gait lab data and estimated values. Despite ignoring the dynamic behaviors or interactions between the prosthesis and the patient, the estimated vertical component of the GRF was highly accurate. Results of this type support the idea that during single stance support and slow walking, a static assumption is sufficient for the estimation of the vertical component of the GRF and ZMP in an instrumented powered prosthesis.

The small magnitudes of the horizontal components of the GRF, compared to the magnitude of the vertical component, reduce the influence of these values in the estimation of the ZMP location.

VI. CONCLUSION

In this paper a method to estimate the GRF and ZMP in autonomous powered ankle-foot prosthesis is described. A clinical assessment to evaluate the accuracy of the estimation methods is performed. The accuracy of the estimated GRF and ZMP suggests that the parameters from the estimation method can be used in the development of improved biomimetic control strategies for biologically realistic lower-limb assistive devices, and thus, improve amputee ambulation.

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