Journal of Biomechanics 44 (2011) 1673–1677

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Trajectory of the center of rotation in non-articulated energy storage and return prosthetic feet

Andrew Sawers [a,b], Michael E. Hahn [a,c,*]

[a] Department of Veterans Affairs (VA), Rehabilitation Research and Development Center of Excellence for Limb Loss Prevention and Prosthetic Engineering, VA Puget Sound, 1660 S. Columbian Way, Seattle, WA 98108, USA
[b] University of Washington, Department of Rehabilitation Medicine, 1959 NE Pacific Street, Box 356490, Seattle, WA 98195, USA
[c] University of Washington, Department of Mechanical Engineering, 4518 University Way Northeast, Seattle, WA 98105, USA

ARTICLE INFO

Article history:
Accepted 24 March 2011

Keywords:
Prosthetic feet
Helical axis
Link-segment model
Amputee
Artificial limb

ABSTRACT

Non-articulated energy storage and return prosthetic feet lack any true articulation or obvious point of rotation. This makes it difficult to select a joint center about which to estimate their kinetics. Despite this absence of any clear point of rotation, methods for estimating the kinetic performance of this class of prosthetic feet typically assume that they possess a fixed center of rotation and that its location is well approximated by the position of the contralateral lateral malleolus. To evaluate the validity of this assumption we used a finite helical axis approach to determine the position of the center of rotation in the sagittal plane for a series of non-articulated energy storage and return prosthetic feet. We found that over the course of stance phase, the sagittal finite helical axis position diverged markedly from the typically assumed fixed axis location. These results suggest that researchers may need to review center of rotation assumptions when assessing prosthetic foot kinetics, while clinicians may need to reconsider the criteria by which they prescribe these prosthetic feet.

1. Introduction

In the study of lower limb prosthetics, joint kinetics are used to analyze the performance of prosthetic components and motor control strategies adopted by individuals with lower limb loss (Winter and Sienko, 1988; Gitter et al., 1991; Underwood et al., 2004). These kinetic descriptions of movement are calculated using pre-defined joint axes (Winter, 2009), whose positions require accurate approximation for estimating hip (Delp and Maloney, 1993; Stagni et al., 2000), and knee joint moments (Holden and Stanhope, 1998). Analyses have rarely been performed on joint axis localization (Rusaw and Ramstrand, 2010) and its importance for estimating kinetic parameters in prosthetic componentry (Prince et al., 1994; Geil et al., 2000; Miller and Childress, 2005).

Non-articulated energy storage and return (NA-ESR) prosthetic feet are typically constructed from carbon fiber composite, have a "J" shape design and lack a clearly-defined axis of rotation. To facilitate comparison with the natural foot–ankle complex and simplify the required calculations, typical assessments of NA-ESR prosthetic foot performance have used constrained link-segment models which assume that the ankle axis of rotation is approximated by the lateral malleolus position (Gitter et al., 1991; Barr et al., 1992; Powers et al., 1998; Underwood et al., 2004; Su et al., 2008; Supan et al., 2010) and behaves as a fixed hinge. In NA-ESR prosthetic feet this assumption is problematic as no true "ankle" articulation exists, the extent to which the joint center remains fixed is unknown, and its approximation by the lateral malleolus has been questioned (Rusaw and Ramstrand, 2010).

To account for any uncertainty in joint power and energy estimates caused by the movement or mis-location of the axis of rotation in NA-ESR prosthetic feet, several groups have incorporated translational power terms into their inverse dynamic analyses (Prince et al., 1994; Geil et al., 2000) on the basis of work in the anatomical foot–ankle (Buczek et al., 1994). However, the extent to which the inclusion of translational power terms can account for joint center mis-location and/or true joint translations in NA-ESR prosthetic feet has not been established.

In light of the limited information regarding the position of the axis of rotation among NA-ESR prosthetic feet, the objective for the current study was to identify the sagittal plane center of rotation position among a series of NA-ESR prosthetic feet in order to assess the appropriateness of continued use of a fixed joint center in the kinetic analysis of NA-ESR prosthetic feet.

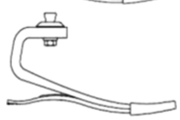
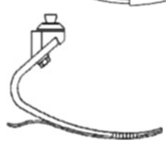
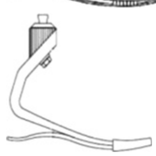
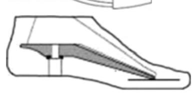
2. Methods

Five NA-ESR prosthetic feet (Table 1) were assessed on one unilateral transtibial amputee (Table 2). Each foot was chosen specifically for the participant

* Corresponding author at: Department of Veterans Affairs (VA), Rehabilitation Research and Development Center of Excellence for Limb Loss Prevention and Prosthetic Engineering, 1660 S. Columbian Way, Seattle, WA 98108, USA.
Tel.: +1 206 277 6310; fax: +1 206 764 2808.
E-mail address: mehahn@uw.edu (M.E. Hahn).

0021-9290/$ - see front matter Published by Elsevier Ltd.
doi:10.1016/j.jbiomech.2011.03.028

Table 1
Prosthetic feet tested.

Foot number	Design	Type	Name (*manufacturer*)	Mass	Material
1[∞]		NA-ESR	Pacifica *(Freedom Innovations; Irvine, CA)*	469 g	Carbon fiber
2[∞]		NA-ESR	Highlander *(Freedom Innovations; Irvine, CA)*	530 g	Carbon fiber
3[∞]		NA-ESR	Senator *(Freedom Innovations; Irvine, CA)*	549 g	Carbon fiber
4[∞]		NA-ESR	Sierra *(Freedom Innovations; Irvine, CA)*	563 g	Carbon fiber
5[α]		NA-ESR	Seattle litefoot *(Seattle Systems; Puolsbo, WA)*	594 g	Delrin

Images modified from http://www.freedom-innovations.com[∞] and http://www.fillauer.com[α]

Table 2
Participant characteristics.

Age	Weight	Gender	Etiology	Time since limb loss	Prescribed prosthesis	Activity level
64 yrs	172 lbs	Male	Traumatic	6.5 yrs	*Socket*: Total Surface Bearing *Interface*: Gel Liner *Suspension*: Pin *Design*: Endoskeletal *Foot*: NA-ESR	MFCL: 3

MFCL: Medicare functional classification level.

in this study based upon their shoe size, body weight and activity level. Each prosthetic foot was initially aligned to the manufacturers' recommendations by a licensed prosthetist. The participant was then allowed time in the lab to become accustomed to each foot condition, during which alignment changes were made to the prosthesis based upon clinical observation and participant reported comfort. The same footwear and prosthetic socket were used for all conditions. Five successful trials per prosthetic foot were collected as the participant walked along a straight 10 m walkway at 1.25 m/s. A trial was considered successful when the prosthetic foot hit the force plate cleanly. All protocols were approved by the Institutional Review Board of the VA Puget Sound Health Care System and the University of Washington. Informed consent was obtained from the participant prior to enrolling in the study.

Sagittal plane center of rotation position was identified in the prosthetic feet using a finite helical axis (FHA) technique (Woltring et al., 1985). The FHA describes the motion between two objects as a rotation about and translation along an axis which can change its position and orientation (Woltring et al., 1985; Blankevoort et al., 1990). This method was selected because of its ability to quantify changes in position of an axis over the course of a movement in situations where that axis is also not easily identified.

Reflective markers, 14 mm in diameter, were placed on the prosthetic shank (wand), as well as the heel, 2nd metatarsal and lateral malleolus of the prosthetic foot. For each of the prosthetic feet, the markers for the heel, 2nd metatarsal and lateral malleolus were placed to match the position of those on the intact contralateral limb. Two additional markers were placed on the medial and lateral sides of the prosthetic pylon for use in calculation of the FHA. Markers on the prosthetic shank were placed in a non-coplanar manner with distribution greater than 10 cm to reduce the potential for error in calculation of the FHA position (Metzger et al., 2010).

Marker coordinate data were collected at 120 Hz using a 12 camera Vicon MX motion system (Vicon Motion Systems, Oxford, UK). After marker trajectories were labeled, a second copy of the raw marker coordinates was made. The first copy of the raw marker coordinates were filtered using a fourth-order Butterworth zero-lag 6 Hz low-pass design (Winter, 2009) in Visual 3D (C-Motion, Germantown, MD). Using custom written MATLAB™ (MathWorks, Natick, MA) code, these filtered marker coordinates were used to calculate shank angle rotation steps of five degrees in the prosthetic limb. This was done to establish a time index of rotation steps over which the FHA would be calculated. The five degree step size was selected in order to minimize errors in the position of the FHA (Woltring et al., 1985), yet maintain resolution. Using a five degree rotation step size resulted in 13–16 rotation steps per stance phase depending on the particular trial.

The FHA was calculated over each shank rotation step using prosthetic shank marker coordinates with respect to the prosthetic foot. These coordinates were obtained from the duplicate copy of the raw marker coordinates and were smoothed using a generalized cross validation quintic spline, strongly advocated in the literature to reduce error in calculation of the FHA (Woltring et al., 1985; Blankevoort et al., 1990; de Lange et al., 1990). Using KineMat (Reinschmidt and van den Bogert, 1997), a MATLAB™ based set of functions, the FHA was calculated from the smoothed marker coordinates. These functions implement a singular value decomposition method (Soderkvist and Wedin, 1993) to obtain 4×4 homogeneous transformation matrices over each rotation step. From these, the parameters of the FHA were then calculated (Spoor and Veldpaus, 1980). We utilized the point where the FHA intersects the sagittal plane of the prosthetic foot coordinate system which was determined from marker coordinates and traditional cross products used to construct local coordinate systems. Differences between the FHA position and the traditionally assumed fixed axis position were examined over stance phase as an ensemble average of five trials per foot. The validity of our FHA point estimate was assessed using a bench top test in which we determined the sagittal plane FHA position between two rigid beams connected by a simple hinge axis. Markers were placed on the lateral aspect of the hinge and the proximal and distal beams to replicate their location on the subject's prosthesis. The FHA positions were then interpolated using a piecewise cubic spline to generate 100 points to plot over stance phase.

3. Results

The validation procedure resulted in an FHA position error range between 0.02 and 2.78 mm in the anterior–posterior direction; between 0.08 and 2.12 mm in the superior–inferior direction.

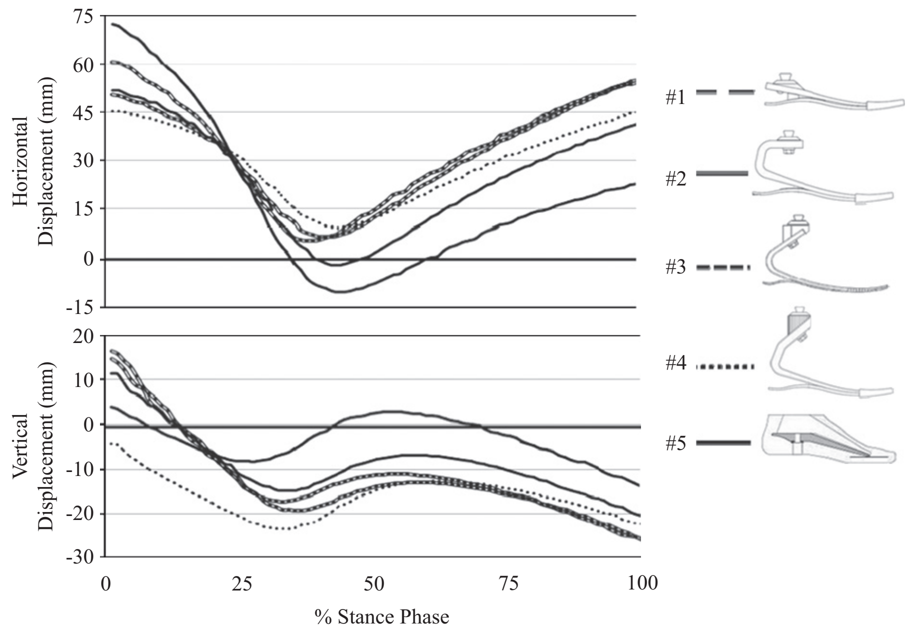


Fig. 1. Position of the finite helical axis in five NA-ESR prosthetic feet relative to the assumed fixed axis position (denoted by the zero line) over stance phase. Positive values on the vertical axis represent anterior or superior positions, while negative values represent posterior or inferior positions. Each foot is represented by an ensemble average of five trials. Data points were calculated over five degrees increments of shank rotation and then interpolated. All feet were tested with their respective cosmetic covers and the subject's preferred footwear.

The RMSE was 1.32 and 1.15 mm for the horizontal and vertical positions, respectively. The error range for the reconstruction of the physical axis marker position was between 0.01 and 0.37 mm in the anterior–posterior direction; between 0.001 and 0.14 mm in the superior–inferior direction. The RMSE was 0.18 and 0.05 mm for the horizontal and vertical positions, respectively.

Calculation of the FHA revealed that sagittal plane position of the center of rotation was not fixed, diverging notably from the assumed center of rotation. Over the stance phase, the FHA position was generally found to be anterior and inferior to this location (Fig. 1).

While point estimates varied between feet, a general pattern emerged for the horizontal trajectory of the FHA during stance phase. In all feet the FHA was at its peak anterior position, 45–74 mm forward of the assumed fixed axis position, after the first 5 degrees of tibial rotation. Over the initial 40–45% of stance the FHA shifted posteriorly by 35–82 mm, moving towards the assumed fixed axis location and reaching its peak posterior position. Over the remaining 55–60% of stance the FHA shifted anteriorly by 32–50 mm, finishing 22–55 mm anterior to the assumed fixed position.

Similarly, the vertical FHA position was found to diverge from the assumed fixed axis location (Fig. 1). In all feet, the FHA began stance phase at its peak superior position, ranging from 22 mm superior to 4 mm inferior to the assumed fixed axis location. Over the initial 30–35% of stance, the FHA shifted inferiorly by 12–33 mm, moving below the assumed fixed axis position. The FHA then briefly shifted superiorly, remaining below the assumed fixed axis location in all but one of the feet. This was followed by a second inferior shift of 10–16 mm over the remaining 40% of stance, resulting in a peak inferior position of 13–26 mm below the assumed fixed axis location.

4. Discussion

The objective of the current work was to locate the position of the center of rotation in a series of NA-ESR prosthetic feet, in order to assess the validity of assuming a fixed joint center for the estimation of their joint kinetics using inverse dynamics. The sagittal plane center of rotation position was found to diverge markedly from the assumed fixed axis location, indicating that assuming a fixed joint center location that matches the lateral malleolus may be unwise.

As anticipated by Czerniecki et al (1991) and Geil et al. (2000), the center of rotation in the sagittal plane was found to behave as a constantly shifting point over the course of stance phase. Recently, Rusaw and Ramstrand (2010) used a functional joint center algorithm (Schwartz and Rozumalski, 2005) to estimate the center of rotation position in a series of commonly prescribed prosthetic feet. While this approach results in a fixed point to represent the center of rotation position over stance phase, they found that across their sample of prosthetic feet, the center of rotation was located anteriorly (range: 28–58 mm) and inferiorly (range: 7–44 mm) to the assumed fixed position, even for a single-axis prosthetic foot (Rusaw and Ramstrand, 2010). Similarly, we found that the FHA was predominantly anterior and inferior to the assumed center of rotation position (Fig. 1). Their results support our finding that the position of the center of rotation in NA-ESR prosthetic feet is not well estimated by the position of the lateral malleolus. Rusaw and Ramstrand (2010) suggest that a benefit of a single point estimated via a functional joint center approach is the ease of application for calculation of joint moments and powers. Based upon the extent to which the FHA was found to displace throughout stance phase in the present study, the use of any fixed joint center in the estimation of prosthetic foot kinetics, regardless of its position, would likely provide very little improvement to the validity of model output. It appears that the center of rotation position among NA-ESR prosthetic feet is not fixed, demonstrating a clear departure from the assumed fixed position, calling into question further use of a fixed joint center regardless of its position.

As a potential solution to improve model validity, it may seem appealing to use the FHA position as the joint center position of a prosthetic to estimate joint kinetics. Unfortunately, it is impractical to implement such an approach in routine inverse dynamic analysis (van den Bogert et al., 2008) and current motion analysis software is unable to account for a moving joint center. As a result, the implementation of such an approach would require custom software that may become cumbersome, computationally expensive and therefore unlikely to be widely adopted. Among the theoretical barriers to implementing a moving FHA position in the estimation of joint kinetics is resolving the violation of rigid body segments if a moving FHA were to act as a segment endpoint. Additionally, if the FHA is not acting as the segment endpoint, the transfer of subsequent reaction forces and moments of force becomes problematic.

Despite not directly calculating joint moments or powers, the results of this study can be used to support the notion that prosthetic foot kinetics are likely overestimated when a fixed center of rotation is used whose position is equivalent to the lateral malleolus (Gitter et al., 1991). For example, the typical internal plantar flexor moment calculated about the traditional position of the fixed joint center during terminal stance would be much reduced if the FHA position were used due to the notably reduced moment arm between the vertical ground reaction force and the joint center. This in turn would likely result in a subsequent decrease in the joint power and therefore work performed by the prosthetic foot over this phase of the gait cycle.

The possibility of such an outcome indicates that NA-ESR prosthetic feet may return less energy to the residual limb than would be expected if a fixed center of rotation position were used. This contradicts a commonly held clinical belief that NA-ESR prosthetic feet should be prescribed on the basis of their ability to return energy during terminal stance. Thus, the rationale for NA-ESR foot prescription and use may require re-examination and the consideration of other factors such as the longer effective foot lengths and near biomimetic roll-over shapes (Hansen et al., 2000) they provide.

It is possible that FHA position during stance phase could be manipulated by prosthetic foot designers to alter the energy storage and return characteristics of prosthetic feet. For example, a prosthetic foot could be designed where the FHA begins stance phase in an anterior position, allowing compliance and energy storage and then moves posteriorly during later stance to provide a greater moment arm at push-off, and a subsequent increase in prosthetic ankle joint power. This notion faces practical challenges but may be worth further exploration.

It is important to note that the outcomes of this investigation reflect an inherent interaction between the prosthetic foot and the shoe worn by the participant. While testing with the shoe replicates the conditions under which these feet will be used, it is possible footwear selection could influence the calculation of the FHA position by slightly altering the motion of the pylon relative to the prosthetic foot, though this seems unlikely.

Use of a single participant in this study limits the external validity of the presented results. Limited accommodation time may also have influenced the results. Calculation of the FHA assumes that the two segments are rigid objects. To minimize the deformable nature of the systems in this study we considered the prosthetic pylon as one segment (assumed to be rigid), and the shoe covering the prosthetic foot as the other segment (considered quasi-rigid since it does not move during stance phase). The quasi-rigid nature of the shoe/foot combination makes the FHA position dependent on marker position. Specific attention was paid to marker placement according to recommendations in the literature (Metzger et al., 2010).

While additional assumptions regarding the link-segment model are commonly made, the aim of this study was to determine the validity of a single model assumption (fixed joint axis). For further review of the impact of additional link-segment model assumptions on assessment of prosthetic componentry the reader is directed elsewhere (Sawers and Hahn, 2010).

Additional research is needed to confirm the findings of this study. Nonetheless, given the sizable displacement demonstrated by the FHA it would appear unwise to continue using a fixed center of rotation that mimics the position of the lateral malleolus on the contralateral limb. Given the impractical nature of implementing the FHA as the joint center in a routine inverse dynamic analysis and the inadequacy of using a fixed point given the magnitude and constant nature of the movement of the FHA, direct measurement via instrumentation of prosthetic feet may provide a gold standard to which all other methods could be compared. In the only reported case to date, the use of standard inverse dynamic methods appears to overestimate energy efficiency when compared to direct measurement (Geil et al., 2000). In addition to improving the validity of kinetic measures of prosthetic foot function, the use of direct measurement may contribute to increasing the validity of proximal joint kinetic estimates, knowledge of which is arguably of greater importance for rehabilitation than additional confirmation of the inefficiency of existing passive prosthetic foot designs.

In conclusion, this study found that the center of rotation position in a series of NA-ESR prosthetic feet was not fixed, and demonstrated notable divergence from the traditionally assumed position throughout stance phase. Considering these results, it is suggested that researchers review center of rotation assumptions prior to biomechanical analysis of NA-ESR prosthetic feet. Further, providers of prosthetic care may need to reconsider the criteria by which they prescribe NA-ESR prosthetic feet based on interpretation of standard link-segment kinetic model outcomes related to prosthetic components.

Conflict of interest statement

Neither of the authors has a potential conflict of interest (e.g., consultancies, stock, and ownership) related to the manuscript or the work it describes.

Acknowledgements

This research was supported in part by a Graduate Student Grant-in-Aid from the American Society of Biomechanics, a Graduate Fellow Research Award from the Orthotic and Prosthetic Education and Research Foundation (OPERF-2009-FA-1), a Research Award from the Center for Orthotics and Prosthetics Learning and Outcomes/Evidence-Based Practice (COPL) and a Center of Excellence grant (A4883C), Department of Veterans Affairs, Rehabilitation Research and Development Service.

References

Barr, A.E., Siegel, K.L., Danoff, J.V., McGarvey 3rd, C.L., Tomasko, A., Sable, I., Stanhope, S.J., 1992. Biomechanical comparison of the energy-storing capabilities of SACH and Carbon Copy II prosthetic feet during the stance phase of gait in a person with below-knee amputation. Phys. Ther. 72, 344–354.
Blankevoort, L., Huiskes, R., de Lange, A., 1990. Helical axes of passive knee joint motions. J. Biomech. 23, 1219–1229.
Buczek, F.L., Kepple, T.M., Siegel, K.L., Stanhope, S.J., 1994. Translational and rotational joint power terms in a six degree-of-freedom model of the normal ankle complex. J. Biomech. 27, 1447–1457.
Czerniecki, J.M., Gitter, A., Munro, C., 1991. Joint moment and muscle power output characteristics of below knee amputees during running: the influence of energy storing prosthetic feet. J. Biomech. 24, 63–75.
de Lange, A., Huiskes, R., Kauer, J.M., 1990. Effects of data smoothing on the reconstruction of helical axis parameters in human joint kinematics. J. Biomech. Eng. 112, 107–113.
Delp, S.L., Maloney, W., 1993. Effects of hip center location on the moment-generating capacity of the muscles. J. Biomech. 26, 485–499.

Geil, M.D., Parnianpour, M., Quesada, P., Berme, N., Simon, S., 2000. Comparison of methods for the calculation of energy storage and return in a dynamic elastic response prosthesis. J. Biomech. 33, 1745–1750.

Gitter, A., Czerniecki, J.M., DeGroot, D.M., 1991. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. Am. J. Phys. Med. Rehabil. 70, 142–148.

Hansen, A.H., Childress, D.S., Knox, E.H., 2000. Prosthetic foot roll-over shapes with implications for alignment of trans-tibial prostheses. Prosthet. Orthot. Int. 24, 205–215.

Holden, J.P., Stanhope, S.J., 1998. The effect of variation in knee center location estimates on net knee joint moments. Gait Posture 7, 1–6.

Metzger, M.F., Faruk Senan, N.A., O'Reilly, O.M., Lotz, J.C., 2010. Minimizing errors associated with calculating the location of the helical axis for spinal motions. J. Biomech 43, 2822–2829.

Miller, L.A., Childress, D.S., 2005. Problems associated with the use of inverse dynamics in prosthetic applications: an example using a polycentric prosthetic knee. Robotica 23, 329–335.

Powers, C.M., Rao, S., Perry, J., 1998. Knee kinetics in trans-tibial amputee gait. Gait Posture 8, 1–7.

Prince, F., Winter, D.A., Sjonnensen, G., 1994. A new technique for the calculation of the energy stored, dissipated, and recovered in different ankle–foot prostheses. IEEE Trans. Rehabil. Eng. 2, 247–255.

Reinschmidt, C., van den Bogert, A.J. 1997. KineMat: A MATLAB toolbox for the reconstruction of spatial marker positions and for the analysis of three-dimensional joint movements. In.

Rusaw, D., Ramstrand, N., 2010. Sagittal plane position of the functional joint centre of prosthetic foot/ankle mechanisms. Clin. Biomech. (Bristol, Avon) 25, 713–720.

Sawers, A.B., Hahn, M.E., 2010. The potential for error with use of inverse dynamic calculations in gait analysis of individuals with lower limb loss: a review of model selection and assumptions. J. Prosth. Orthot. (JPO) 22, 56–61.

Schwartz, M.H., Rozumalski, A., 2005. A new method for estimating joint parameters from motion data. J. Biomech. 38, 107–116.

Soderkvist, I., Wedin, P.A., 1993. Determining the movements of the skeleton using well-configured markers. J. Biomech. 26, 1473–1477.

Spoor, C.W., Veldpaus, F.E., 1980. Rigid body motion calculated from spatial co-ordinates of markers. J. Biomech. 13, 391–393.

Stagni, R., Leardini, A., Cappozzo, A., Grazia Benedetti, M., Cappello, A., 2000. Effects of hip joint centre mislocation on gait analysis results. J. Biomech. 33, 1479–1487.

Su, P.F., Gard, S.A., Lipschutz, R.D., Kuiken, T.A., 2008. Differences in gait characteristics between persons with bilateral transtibial amputations, due to peripheral vascular disease and trauma, and able-bodied ambulators. Arch. Phys. Med. Rehabil. 89, 1386–1394.

Supan, T., Lebiedowska, M., Dodson, R., Verhulst, S., Dufour, M., 2010. The effect of a Talux® prosthetic foot on gait parameters and limb loading of nonvascular transtibial amputees. J. Prosthet. Orthot. 22, 43–52.

Underwood, H.A., Tokuno, C.D., Eng, J.J., 2004. A comparison of two prosthetic feet on the multi-joint and multi-plane kinetic gait compensations in individuals with a unilateral trans-tibial amputation. Clin. Biomech. (Bristol, Avon) 19, 609–616.

van den Bogert, A.J., Reinschmidt, C., Lundberg, A., 2008. Helical axes of skeletal knee joint motion during running. J. Biomech. 41, 1632–1638.

Winter, D.A., 2009. Biomechanics and motor control of human movement, 4th ed. Wiley, Hoboken, N.J..

Winter, D.A., Sienko, S.E., 1988. Biomechanics of below-knee amputee gait. J. Biomech. 21, 361–367.

Woltring, H.J., Huiskes, R., de Lange, A., Veldpaus, F.E., 1985. Finite centroid and helical axis estimation from noisy landmark measurements in the study of human joint kinematics. J. Biomech. 18, 379–389.