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Effect of alignment changes on socket reaction moments while walking in transtibial prostheses with energy storage and return feet

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Abstract

Background—Energy storage and return feet are designed for active amputees. However, little is known about the socket reaction moments in transtibial prostheses with energy storage and return feet. The aim of this study was to investigate the effect of alignment changes on the socket reaction moments during gait while using the energy storage and return feet.

Methods—A Smart Pyramid™ was used to measure the socket reaction moments in 10 subjects with transtibial prostheses while walking under 25 alignment conditions, including a nominal alignment (as defined by conventional clinical methods), as well as angle malalignments of 2°, 4° and 6° (flexion, extension, abduction, and adduction) and translation malalignments of 5mm, 10mm and 15mm (anterior, posterior, lateral, and medial) referenced from the nominal alignment. The socket reaction moments of the nominal alignment were compared with each malalignment.

Findings—Both coronal and sagittal alignment changes demonstrated systematic effects on the socket reaction moments. In the sagittal plane, angle and translation alignment changes demonstrated significant differences ($P<0.05$) in the minimum moment, the moment at 45% of stance and the maximum moment for some comparisons. In the coronal plane, angle and translation alignment changes demonstrated significant differences ($P<0.05$) in the moment at 30% and 75% of stance for all comparisons.

Interpretation—The alignment may have systematic effects on the socket reaction moments in transtibial prostheses with energy storage and return feet. The socket reaction moments could potentially be a useful biomechanical parameter to evaluate the alignment of the transtibial prostheses.

Keywords

amputation; direct measurement; load; malalignment; kinetics

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Conflict of Interest Statement

Authors of the manuscript (Kobayashi T, Orendurff MS, Rosenbaum-Chou TG and Boone DA) are currently employees of the company that manufactures the Smart Pyramid™ (Currently known as Europa™) used in this study.

1. Introduction

Energy storage and return (ESR) feet are designed for active amputees with prostheses. They have been claimed to assist push-off by releasing energy stored in the flexible keel during mid to late stance. The ratio between stored and returned energy (energy efficiency) depends on the design of the foot, and returned energy is inherently less than absorbed energy as some is lost due to inefficiency of the spring (Czerniecki et al., 1991; Ehara et al., 1993; Geil et al., 2000; Prince et al., 1998). A reduction in stiffness of the ESR foot results in an increase of mid-stance energy storage and late-stance energy return (Fey et al., 2011). How ESR feet may benefit amputees gait and better clinical care has been studied extensively (Gailey et al., 2012; van der Linde et al., 2004).

A variety of ESR feet are currently available in the market. Their characteristics are partly determined by their inherent features, such as the stiffness of the keel or the axis of rotation. However, detailed mechanical characteristics of each ESR foot is a proprietary to each manufacture. It is generally difficult to relate the results of biomechanical analyses of ESR feet to amputee's preference of the foot in the clinic (Hafner et al., 2002). A review paper showed that a number of studies compared the effects of ESR feet to SACH (solid ankle cushion heel) feet, but no robust evidence exists that ESR feet outperform SACH feet (van der Linde et al., 2004). Comparisons of SACH and ESR feet in transtibial prostheses did not demonstrate differences in various clinical assessment parameters, including metabolic cost (Torburn et al., 1995), amputees' preference of feet (Postema et al., 1997b), and temporal-spatial parameters of gait (Perry et al., 1997; Postema et al., 1997a). However, differences in ankle kinematics were reported (Postema et al., 1997a; Schmalz et al., 2002; Torburn et al., 1990).

The alignment of transtibial prostheses is the spatial relationship between the socket and foot. It is tuned by a prosthetist through bench, static and dynamic alignment procedures in the clinic (Ikeda et al., 2012). The effects of alignment changes on amputees have been investigated in gait symmetry (Andres and Stimmel, 1990; Chow et al., 2006; Hannah et al., 1984), socket-residual limb interface pressures or loadings on the limb while walking (Pinzur et al., 1995; Sanders et al., 1998; Seelen et al., 2003; Zhang et al., 1998), and balance or muscular activity while standing (Blumentritt et al., 1999; Isakov et al., 1994). It is anecdotally believed that the alignment is important to maximize the benefit from ESR feet. However, prosthetists have shown a large range of alignment variations as optimal (Zahedi et al., 1986) and amputees' perception of alignment might not be fully reliable (Boone et al., 2012).

Socket reaction moments are conceptually acting around the center of the socket to balance the rotating effect of ground reaction forces during gait (Kobayashi et al., 2013b). They represent the way a residual limb is loaded inside the socket. An external extension moment suggests more loading at proximal-anterior and distal-posterior aspects of the residual limb in the sagittal plane, while an external varus moment suggests more loading at proximal-medial and distal-lateral aspects of the residual limb (Boone et al., 2013). Previous studies demonstrated that socket reaction moments were systematically influenced by alignment changes in transtibial prostheses with SACH feet (Boone et al., 2013; Kobayashi et al., 2013b). A similar effect may be expected in transtibial prostheses with ESR feet.

ESR feet are commonly prescribed for amputees, however; little is known about the effect of alignment changes on the socket reaction moment in prostheses with ESR feet. Therefore, this warrants further work to build more evidence in prosthetic alignment. The aim of this study was to investigate the effect of systematic alignment changes on the socket reaction moments in transtibial prostheses with ESR feet. The hypothesis of the study was that

alignment changes in prostheses with ESR feet would have significant effects ($P<0.05$) on the socket reaction moments.

2. Methods

2.1. Subjects

Ten subjects (4 females / 6 males) aged 50 (11) years old with transtibial prostheses were recruited from the community (Table 1). Their mean height was 1.74 (0.08) m and their mean body mass was 83.6 (17.5) kg. All subjects were users of ESR feet in their daily life. Nine subjects had an amputation because of trauma, while the other subject had an amputation due to peripheral vascular disease. The ESR feet worn by the subjects included Multiflex Foot (Endolite, Miamisburg, OH, USA), Flex Walk (Ossur, Foothill Ranch, CA, USA), Seattle Light Foot (Trulife, Poulsbo, WA, USA), Seattle Voyager Foot (Trulife, Poulsbo, WA, USA), Seattle Carbon Light Foot (Trulife, Poulsbo, WA, USA), Cadence HP (Trulife, Poulsbo, WA, USA), Seattle Catalyst Foot (Trulife, Poulsbo, WA, USA), Century 22 adjustable heel height foot (Century XXII Innovations, Jackson, MI, USA), and Renegade LP (Freedom Innovations, Irvine, CA, USA). This study was approved by the institutional review board governing the institution, and informed consent was obtained from each subject.

2.2. Instrument

Measurement of the socket reaction moments was conducted using an instrumented prosthetic pyramid adaptor: Smart Pyramid™ (Height: 3.22 cm; Mass: 0.14 kg) (Orthocare Innovations, Mountlake Terrace, WA, US) (Figure 1). The sagittal and coronal socket reaction moments are measured by strain gauges located in the anteroposterior and mediolateral beams inside the Smart Pyramid. Signal conditioning, temperature compensation, analogue-to-digital conversion, digital-to-analogue conversion and a serial communication bus are included in a printed circuit board of the Smart Pyramid. The Smart Pyramid can be used in either an upward or downward pyramid configuration depending on the type of the socket adaptor. The sampling frequency of the Smart Pyramid is 100Hz, and the collected data are sent to the host computer wirelessly via Bluetooth for further analyses.

Prior to the experiment, the calibration of the Smart Pyramid was performed by applying an increasing series of known moments from 0 to 109.205 Nm (0, 7.233 Nm, 19.205 Nm, 28.184 Nm, 37.163 Nm, 48.022 Nm, 58.219 Nm, 68.416 Nm, 78.614 Nm, 88.811 Nm, 99.008 Nm and 109.205 Nm) at set angular displacement in the transverse plane to the Smart Pyramid from the anterior 0° position (16 orientations including: 0°, 22.5°, 45°, 67.5°, 90°, 112.5°, 135°, 157.5°, 180°, 202.5°, 225°, 247.5°, 270°, 292.5°, 315°, and 337.5°). The moment output was highly linear with coefficients of determination of $R^2=0.998$ for the sagittal moment and $R^2=0.996$ for the coronal moment. The measured sagittal and coronal moments showed a root mean square error of 2.08 % and 2.80 %, respectively (Kobayashi et al., 2013a).

Angle alignment changes were monitored using Biaxial MEMS tilt sensors (CXT-02, Crossbow Technologies, San Jose, USA), while translation alignment changes were monitored using mm scales on the translation adjuster (10A40/A, F.G. Streifeneder KG, Emmering, Germany) (Boone et al., 2013) (Figure 1).

2.3. Protocol

The Smart Pyramid was attached at the bottom of each amputee's prosthetic socket used in daily life (Figure 1). Alignment of the prosthesis was tuned to the satisfaction of the prosthetist and the subject using classical alignment techniques consistent with those used

by the typical prosthetist in a clinical setting. Angle alignment changes of 2°, 4° and 6° (flexion, extension, abduction, and adduction) and translation alignment changes of 5 mm, 10 mm and 15 mm (anterior, posterior, lateral, and medial) were induced in a random order from the nominally aligned condition. The alignment of the transtibial prosthesis is described in Figure 2. The alignment was described by moving the position of the foot relative to the fixed socket position. Including the nominally aligned condition, there were 25 alignment conditions tested in this study. Immediately after the alignment was adjusted, the participant was instructed to walk down a 10-meter path at a self-selected walking speed. This is in accordance to the current clinical practice of alignment tuning process.

2.4. Data analysis

The socket reaction moments were interpolated with a cubic spline function, and the stance phase was normalized to 100% in 1% increments. The socket reaction moments of 3 steps in each trial were normalized to body mass (Nm/kg) and averaged. The following moment parameters were extracted based on the previous studies (Boone et al., 2013; Kobayashi et al., 2012). Maximum moment, minimum moment, and moment at 45% of stance phase ($M_{\max}$, $M_{\min}$ and M_{45}) were extracted from the sagittal moment, while the moment at 30% of stance phase and the moment at 75% of stance phase (M_{30} and M_{75}) were extracted from the coronal moment. The socket reaction moments are described in Figure 3. An external valgus moment in the coronal plane and an external extension moment in the sagittal plane were defined as positive.

2.5. Statistical analysis

Normality and homogeneity of variance for all the moment parameters and cadence were tested by Bartlett's test. The nominally aligned condition was compared to each of the malaligned conditions (i.e. Aligned vs 2°, 4° and 6° of malalignments; Aligned vs 5 mm, 10 mm and 15 mm of malalignments) for the moment parameters and cadence utilizing 2-tailed paired t-tests with Stata/IC 11.1 (STATA Corp, College Station, USA). No statistical comparisons were performed between malaligned conditions. The influence of sagittal plane angle and translation alignment changes was analyzed in the sagittal plane, while that of coronal plane was analyzed in the coronal plane. Means, standard deviations and 95% confidence intervals of each moment parameter and cadence were calculated. Statistical significance was set at $P = 0.05$. P -values were adjusted for multiple comparisons using the Tukey-Ciminera-Heyse procedure (Sankoh et al., 1997; Tukey et al., 1985).

3. Results

Alignment changes induced systematic and in some cases statistically significant effects on the socket reaction moments. The Bartlett's test confirmed that the socket reaction moment parameters and cadence data were normally distributed.

3.1. Effect of sagittal alignment changes on the sagittal moments

Table 2 shows the means, standard deviations and 95% confidence intervals of sagittal socket reaction moment parameters ($M_{\min}$, M_{45} and $M_{\max}$) in response to the sagittal plane alignment changes. External extension moments and external flexion moments were defined as positive and negative, respectively. Figure 4 shows the effect of alignment changes on the normalized sagittal socket reaction moments during stance. Both angle and translation alignment changes demonstrated significant differences from the nominally aligned condition in the minimum moment, moment at 45% of stance and maximum moment for some comparisons. A more systematic effect was observed in translation alignment changes in comparison to angle alignment changes.

3.2. Effect of coronal alignment changes on the coronal moments

Table 2 shows the means, standard deviations and 95% confidence intervals of the coronal socket reaction moment parameters (M_{30} and M_{75}) in response to the coronal plane alignment changes. External valgus moments and external varus moments were defined as positive and negative, respectively. Figure 5 shows the effect of alignment changes on the normalized coronal socket reaction moments during stance. Both angle and translation alignment changes demonstrated significant differences from the nominally aligned condition in the socket reaction moments measured at 30% and 75% of stance for all comparisons.

3.3. Effect of alignment changes on cadence

Table 3 shows the means, standard deviations and 95% confidence intervals of cadence in response to the sagittal and coronal plane alignment changes. No significant differences were found between the nominally aligned and malaligned conditions for all comparisons.

4. Discussion

This study demonstrated that alignment changes had systematic effects on the socket reaction moments in transtibial prostheses with ESR feet. The pattern of changes in the socket reaction moments in response to the alignment changes was similar to SACH feet in the sagittal and coronal planes (Boone et al., 2013). The previous work tested the malalignment by 3° and 6° in angles and 5mm and 10mm in translations, while this study investigated malalignment by 2°, 4° and 6° in angles and 5mm, 10mm and 15mm in translations. This study demonstrated that significant effects on the socket reaction moments occur by 2° of malalignment. For instance, extension by 2° induced a statistically significant reduction in the maximum moment by 0.051 Nm/kg in the sagittal plane. However, the minimum clinically important difference (smallest improvement considered worthwhile by an amputee) for the socket reaction moment is unknown and requires a further investigation (Copay et al., 2007).

A flexion malalignment tended to induce a decrease (a decrease in the absolute value) in minimum moment and moment at 45% stance as well as an increase in maximum moment. A posterior translation malalignment of the socket tended to induce a decrease in minimum moment as well as an increase in moment at 45% stance and maximum moment in the sagittal plane. Generally under the nominally aligned condition, the sagittal moment should initially demonstrate an external flexion moment, which is measured as a negative moment value at early stance (0-30% of stance). This is followed by an external extension moment, which is measured as a positive moment value. Extension of the socket by 4° and 6° induced a significant increase in the socket reaction moments at 45% of stance. This may relate to the amputee's self reported perception of "walking uphill". Prosthetists typically flex the socket or dorsiflex the foot in this case. Flexion of the socket demonstrated a reduction in the socket reaction moment at 45% of stance. This may justify the anecdotal approach of the prosthetists to alleviate the amputee's perception of "walking uphill". An increase in the maximum moment means increased loading at proximal-anterior and distal-posterior aspects of the residual limb. An increased negative value in the minimum moment means increased loading at distal-anterior and proximal-posterior aspects of the residual limb. The maximum moment may be related to pain or discomfort at the patella tendon region and the minimum moment may be related to those at the end of the tibia region.

Generally under the nominal alignment, the coronal moment should initially demonstrate an external valgus moment, which is measured as a positive moment value at early stance (0-20% of stance). This is followed by an external varus moment, which is measured as a

negative moment value. In the coronal plane, the magnitude and direction of the socket reaction moments were significantly altered by changing alignments. Adduction and medial translation of the socket induced an increase in the valgus moment in the coronal plane. Bench alignment of the transtibial prosthesis is done such that a slight external varus moment is generated during stance for stability. The results of this study suggested that this varus moment could be compromised by malalignments due to excessive adduction or medial translation of the socket. The proximal-medial and distal-lateral aspects of the residual limb have more loading when the varus moment is dominant. The proximal-lateral and distal-medial aspects of the residual limb have more loading when the valgus moment is dominant. The varus moment may be related to pain or discomfort around the distal fibula and the valgus moment may be related to those around the proximal fibula.

The mean maximum moment (0.830 (0.099) Nm/kg) and the mean minimum moment (−0.180 (0.136) Nm/kg) of ESR feet in the sagittal plane for the nominally aligned condition (Table 2) were larger than the SACH feet (0.719 (0.177) Nm/kg and −0.147 (0.117) Nm/kg, respectively) reported in the earlier work (Boone et al., 2013). The maximum and minimum sagittal moment reported in a case series study by direct measurement using a JR3 sensor in transtibial prostheses with ESR feet ranged from 0.609 to 1.109 Nm/kg and from −0.191 to −0.281 Nm/kg, respectively (Neumann et al., 2012). The mean cadence of participants with ESR feet in this study compared to those with SACH feet in the earlier work (Kobayashi et al., 2012) at the nominal alignment was 104 (12) steps/min and 109 (16) steps/min, respectively (Table 3). These results suggest that the range of the socket reaction moments measured in the current study was reasonable, and the cause for the higher socket reaction moments measured in ESR feet may not be explained by temporal-spatial parameters such as cadence or gait speed.

The socket reaction moments may be influenced by mechanical design of prosthetic feet and amputees gait patterns. Each amputee had a similar pattern to their socket reaction moments with the alignment changes made. Although each amputee may have some unique characteristics, and there is step-to-step variability, the patterns are all recognizable and reasonably consistent. The fact that changing the angle and translation alignment produced systematic changes on the socket reaction moments in one direction or another suggests that the nominally aligned position was at the center of these perturbations for the group as a whole. Nine types of ESR feet were used in this study and their mechanical design characteristics are different from each other. Therefore, the results of this study represent the general effects of alignment changes on the socket reaction moments in ESR feet. Observed variability in the data may be potentially affected by variability of mechanical property of each ESR foot and gait characteristics of each amputee. A further study is required to investigate what design factors of the feet, such as stiffness, hysteresis and shape of the keel may affect the socket reaction moments (Fey et al., 2011; van Jaarsveld et al., 1990).

The differences in sagittal ankle power patterns during walking are likely to be a much more sensitive discriminator for different types of prosthetic feet than the sagittal moment (Geil et al., 1999; Geil et al., 2000). The stiffness of the forefoot portion of the keel determines the amount of displacement, bending toward dorsiflexion as midstance progresses, and recoiling toward plantarflexion in late stance and pre-swing (Prince et al., 1998). Dorsiflexion bending creates power absorption and recoiling toward plantarflexion creates power generation. Differences in stiffness of the forefoot portion of the prosthetic foot keel that distinguish ESR feet from SACH feet, and that distinguish specific ESR feet from one another are not likely to have a dramatic effect on the sagittal moments measured in this study. The differences are primarily related to displacement with a distal load (moment), and the loading pattern is more closely associated with the alignment of the prosthetics components.

It is unlikely that there is one alignment that is optimal for all transtibial amputees. Biomechanically optimal alignment should enable amputees to have the most energy efficient gait possible. However, defining a single optimal alignment is beyond the scope of this study. It may not be one exact position. It might be an acceptable range; more narrow or broad depending on the function and perception of the amputee. Some individuals may respond differently to specific angle or translation alignment changes, but there is a generalized response. Increasing and decreasing the angle or translation produced systematic changes in the socket reaction moment in a positive direction or a negative direction. This is an indication that the nominally aligned position was at the center of these perturbations for the group as a whole. This suggests that the socket reaction moment could be one metric of the biomechanical effect of differences in the prosthetic alignment.

Prosthetic alignment is currently tuned visually, both by viewing the angle of the socket-pylon-foot system and by visual gait observation of the patient's gait with input from the amputee. Currently, this optimization is performed without the benefit of any kinetic analysis. Each individual may tolerate a different range of possible alignments and some individuals may be able to mask the effect of malalignment at the cost of greater and inappropriate loads on the residual limb as well as the contralateral limb and additional energy expenditure. A visual gait analysis has substantial limits on accuracy and most patients have a limited ability to perceive and communicate alignment changes (Boone et al., 2012). Previous research has suggested that with only visual observation of gait and amputee's feedback there is a broad range of acceptable alignments that satisfy both the prosthetist and amputee (Zahedi et al., 1986). One well-validated solution is to have full-body 3D kinematic and kinetic data to show the effect of changes in alignment. Many different metrics can be affected, and kinematics and kinetics may interact. However, a computerized gait analysis using a 3D motion analysis system is prohibitively complex, expensive, very time consuming and difficult for clinicians to interpret and incorporate into their daily practice. Therefore, the direct measurement of kinetic parameters of prosthetic gait (Frossard et al., 2003; Neumann et al., 2012; Sanders et al., 1997) is more feasible in the clinic than indirect kinetic measurement using inverse dynamics that requires a motion analysis laboratory (Dumas et al., 2009; Stephenson and Seedhom, 2002). Since problems associated with amputation are often related to abnormal loading, not abnormal angles *per se*, it is reasonable to look at the kinetic parameters, such as the socket reaction moment in order to make judgments about alignment. Additional measures, such as socket-residual limb interface pressures, oxygen consumptions or whole body (center of mass) movement may be valuable to investigate the effect of alignment on amputees.

It is anecdotally known that alignment plays an important role in prosthetic function. Even with perfect alignment the joints and tissues of an individual with limb loss are likely to be subjected to abnormal forces, shears, moments and impulses. The goal of lower-limb prosthetic fittings is to minimize these abnormal forces and permit the best possible gait and ambulatory functional performance while avoiding deleterious affects. Abnormal loading on the residual limb may be attributed to gait compensations that arise from the inability of prosthetic components to mimic biological functions. To what extent the alignment will contribute to improve residual limb loading is beyond the scope of this study. This study only investigated the moment. A recent case series study suggested that force might be a predictor of pressure at the distal tibia, gastrocnemius and popliteal regions during stance, while the moment might be the predictor of pressure at the patella tendon region during forefoot loading (Neumann et al., 2013). The relationship among the alignment, forces, moments and pressures should be explored further.

While long-term adaptation to a particular alignment might occur, this study was intended to investigate the immediate effects of alignment changes. This was done in a manner similar

to a current clinical practice: adjust and observe in an iterative manner until both a prosthetist and an amputee are satisfied with the alignment. The effect of accommodation of alignment on the socket reaction moments should be investigated in a further study. The effect of alignment was investigated under 25 alignment conditions in a randomized order. Learning and fatigue may have affected the results of the study. The socket reaction moment may also be influenced by gait speed considering its effect on lower limb joint moments (Goldberg and Stanhope, 2013). Although gait speed was not measured, cadence did not reveal any statistical differences among the alignment conditions tested in this study. It is unlikely that gait speed affected the results. A future study should investigate the relationship between gait speed and the socket reaction moments.

5. Conclusions

The results of this study suggested that alignment changes had a systematic effect on the socket reaction moments in the sagittal and coronal planes in transtibial prostheses with ESR feet. The socket reaction moments could potentially be a useful biomechanical parameter to evaluate the alignment of the transtibial prostheses with ESR feet. A future study should investigate the effect of the alignment on the long-term clinical care of the amputees with lower-limb prostheses.

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Smart Pyramid

Experimental Setup

Figure 1.
Smart Pyramid™ and experimental setup.

(A) Angle alignment changes

(B) Translation alignment changes

Figure 2.
Description of prosthetic alignment: (A) Angle alignment changes, (B) Translation alignment changes.

Figure 3.
Description of socket reaction moments.

A. Sagittal Angular Socket Changes

B. Sagittal Translation Socket Changes

Figure 4.

Normalized socket reaction moment (Nm/kg) in response to (A) sagittal angle changes and (B) sagittal translation changes. Mean (◆) and 95% confidence interval are provided at minimum moment, moment at 45% stance and maximum moment. Maximum and minimum values on the graphical display of the moment curves may not match the maximum or minimum values extracted for statistical hypothesis testing [small point and whisker graphs]. The full moment curves have temporal variability in the peak timing of the maximum and minimum values whereas the hypothesis testing was performed on the mean calculated from each individual's peak value regardless of timing.

Abbreviation: Ant, anterior; Ext, extension; Flex, flexion; Post, posterior

A. Coronal Angular Socket Changes

B. Coronal Translation Socket Changes

Figure 5.

Normalized socket reaction moment (Nm/kg) in response to (A) coronal angular changes and (B) coronal translation changes. Mean (◆) and 95% confidence interval are provided at moment at 30% stance and moment at 75% stance.

Abbreviation: Abd, abduction; Add, Adduction; Lat, lateral; Med, medial

Table 1

Demographic data of the subjects

Subject	Gender	Age	Height (m)	Mass (kg)	Residual limb length (cm)	Years since amputation	Amputated side	Foot
1	Male	46	1.85	85	18	6	Left	Multiflex Foot
2	Male	53	1.85	93	13	9	Left	Flex Walk
3	Male	70	1.78	83	16	48	Left	Seattle Light Foot
4	Female	50	1.64	63	13	18	Right	Seattle Voyager Foot
5	Female	64	1.75	64	13	32	Right	Seattle Carbon Light
6	Male	45	1.83	101	14	20	Right	Cadence HP
7	Male	35	1.73	75	13	15	Left	Seattle Catalyst Foot
8	Female	49	1.65	112	14	9	Left	Century 22 AHHF
9	Female	35	1.63	62	16	12	Left	Renegade LP
10	Male	51	1.73	98	20	1	Right	Seattle Light Foot

Table 2

Effect of sagittal and coronal alignment changes on normalized socket reaction moments

Alignment Perturbations	Mmin (Nm/kg)			M45 (Nm/kg)			Mmax (Nm/kg)		
	Mean (SD)	95% CI		Mean (SD)	95% CI		Mean (SD)	95% CI	
Sagittal Angle	6°Ext	-0.226 (0.121)	-0.313, -0.139	0.319 (0.104) ^(†)	0.245, 0.393		0.698 (0.119) ^(†)	0.613, 0.783	
	4°Ext	-0.214 (0.134)	-0.310, -0.118	0.306 (0.137) ^(*)	0.208, 0.404		0.750 (0.107) ^(†)	0.673, 0.827	
	2° Ext	-0.212 (0.135)	-0.309, -0.115	0.277 (0.150)	0.170, 0.384		0.779 (0.121) ^(†)	0.693, 0.865	
	Aligned	-0.180 (0.136)	-0.278, -0.082	0.245 (0.155)	0.134, 0.356		0.830 (0.099)	0.759, 0.901	
	2° Ext	-0.155 (0.141)	-0.256, -0.054	0.231 (0.141)	0.130, 0.332		0.837 (0.124)	0.748, 0.926	
	4° Flex	-0.145 (0.166)	-0.264, -0.026	0.202 (0.135)	0.106, 0.298		0.838 (0.127)	0.747, 0.929	
	6° Flex	-0.107 (0.142) ^(†)	-0.209, -0.005	0.163 (0.197)	0.022, 0.304		0.850 (0.138)	0.751, 0.949	
	15mm Ant	-0.274 (0.157) ^(†)	-0.387, -0.161	0.157 (0.150) ^(*)	0.050, 0.264		0.694 (0.090) ^(†)	0.630, 0.759	
	10mm Ant	-0.243 (0.144) ^(†)	-0.346, -0.140	0.182 (0.145)	0.078, 0.286		0.712 (0.088) ^(†)	0.650, 0.775	
	5mm Ant	-0.219 (0.131) ^(†)	-0.312, -0.126	0.240 (0.120)	0.154, 0.326		0.742 (0.102) ^(†)	0.669, 0.815	
Sagittal Translation	Aligned	-0.180 (0.136)	-0.278, -0.082	0.245 (0.155)	0.134, 0.356		0.830 (0.099)	0.759, 0.901	
	5mm Post	-0.159 (0.140) ^(*)	-0.259, -0.059	0.298 (0.149) ^(†)	0.191, 0.405		0.855 (0.119)	0.770, 0.940	
	10mm Post	-0.138 (0.138) ^(†)	-0.236, -0.040	0.318 (0.165) ^(†)	0.200, 0.436		0.877 (0.134)	0.781, 0.973	
	15mm Post	-0.113 (0.136) ^(†)	-0.210, -0.016	0.363 (0.143) ^(†)	0.261, 0.465		0.905 (0.134) ^(*)	0.809, 1.001	
Alignment Perturbations	M30 (Nm/kg)			M75 (Nm/kg)					
	Mean (SD)	95% CI		Mean (SD)	95% CI				
Coronal Angle	6° Abd	-0.246 (0.126) ^(†)	-0.336, -0.156	-0.208 (0.092) ^(†)	-0.274, -0.142				
	4° Abd	-0.193 (0.087) ^(†)	-0.255, -0.131	-0.148 (0.088) ^(†)	-0.211, -0.085				
	2° Abd	-0.149 (0.076) ^(†)	-0.204, -0.094	-0.104 (0.089) ^(†)	-0.168, -0.040				
	Aligned	-0.081 (0.064)	-0.127, -0.352	-0.046 (0.082)	-0.105, 0.013				
	2° Add	-0.023 (0.056) ^(†)	-0.063, 0.017	-0.001 (0.088) ^(†)	-0.064, 0.062				
	4° Add	0.031 (0.056) ^(†)	-0.009, 0.071	0.054 (0.101) ^(†)	-0.018, 0.126				

Alignment Perturbations	M30 (Nm/kg)			M75 (Nm/kg)		
	Mean (SD)	95% CI		Mean (SD)	95% CI	
Coronal Translation	6° Add	0.084 (0.061) ^(†)	0.040, 0.128	0.108 (0.110) ^(†)	0.030, 0.186	
	15mm Lat	-0.225 (0.093) ^(†)	-0.291, -0.159	-0.187 (0.091) ^(†)	-0.252, -0.122	
	10mm Lat	-0.184 (0.092) ^(†)	-0.249, -0.119	-0.129 (0.090) ^(†)	-0.194, -0.064	
	5mm Lat	-0.130 (0.081) ^(†)	-0.188, -0.072	-0.101 (0.095) ^(†)	-0.169, -0.033	
	Aligned	-0.081 (0.064)	-0.127, -0.035	-0.046 (0.082)	-0.105, 0.013	
	5mm Med	-0.043 (0.075) ^(†)	-0.097, 0.011	-0.018 (0.080) ^(†)	-0.075, 0.039	
	10mm Med	0.009 (0.065) ^(†)	-0.038, 0.056	0.023 (0.088) ^(†)	-0.040, 0.086	
	15mm Med	0.057 (0.071) ^(†)	0.006, 0.108	0.068 (0.085) ^(†)	0.007, 0.129	

Abbreviations: 95% CI, 95% confidence interval; Abd, abduction; Add, adduction; Flex, flexion; Lat, lateral; M_{min}, minimum socket reaction moment; M_{max}, maximum socket reaction moment; M30, socket reaction moment at 30% of stance phase; M45, socket reaction moment at 45% of stance phase; M75, socket reaction moment at 75% of stance phase; Med, medial; Post, posterior; SD, standard deviation

^(*) An asterisk indicates significant difference at $P<0.05$ from the aligned condition, while a cross

^(†) indicates significant difference at $P<0.01$ from the nominally aligned condition.

Table 3

Effect of sagittal and coronal alignment changes on cadence

Alignment Perturbations		Cadence (steps/min)		Alignment Perturbations		Cadence (steps/min)	
		Mean (SD)	95% CI			Mean (SD)	95% CI
Sagittal Angle	6°Ext	105 (12)	96, 113	6° Abd		102 (11)	95, 110
	4°Ext	105 (11)	96, 113	4° Abd		105 (13)	95, 114
	2° Ext	105 (12)	96, 114	2° Abd		105 (12)	96, 114
	Aligned	104 (12)	95, 112	Aligned		104 (12)	95, 112
	2° Flex	104 (14)	94, 114	2° Add		104 (12)	95, 113
	4° Flex	105 (12)	96, 113	4° Add		105 (12)	97, 113
Sagittal Translation	6° Flex	104 (11)	96, 112	6° Add		104 (12)	95, 113
	15mm Ant	104 (13)	94, 113	15mm Lat		103 (12)	94, 112
	10mm Ant	104 (13)	95, 114	10mm Lat		104 (10)	97, 112
	5mm Ant	104 (12)	95, 112	5mm Lat		104 (12)	96, 112
	Aligned	104 (12)	95, 112	Aligned		104 (12)	95, 112
	5mm Post	104 (13)	95, 113	5mm Med		103 (12)	95, 111
	10mm Post	104 (12)	95, 112	10mm Med		104 (12)	95, 112
	15mm Post	104 (12)	96, 113	15mm Med		103 (12)	95, 112

Abbreviations: 95% CI, 95% confidence interval; Abd, abduction; Add, adduction; Ant, anterior; Ext, extension; Flex, flexion; Lat, lateral; Med, medial; Post, posterior; SD, standard deviation