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Foot trajectories and loading rates in a transfemoral amputee for six different commercial prosthetic knees: An indication of adaptability

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ABSTRACT

Background: The relationship between the functional loading rate and heel velocities was assessed in an active unilateral transfemoral amputee (UTFA) for adaptation to six different commercial prosthetic knees.
Objective: To Investigate the short-term process of adaptability for UTFA for two types of prosthetic knees were evaluated, based on the correlation between heel vertical velocity and transient loading rate.
Methods: The loading rate was calculated from the slope of ground reaction forces (GRF) and the corresponding time. The heel velocities and GRF were obtained by a motion analysis system.
Results: Biomechanical adaptation was evident following a short period of prosthetic knee use based upon the mean transient impact (loading rate) and the heel vertical velocity in slow, normal and fast walking. Trend lines of transient impact versus vertical heel velocity for a set of actively controlled variable damping (microprocessor) and mechanically passive prosthetic knees were all negatively correlated, except for an amputated leg during normal pace and healthy leg during fast pace. For an amputee to adapt well to a prescribed prosthesis excellent coordination between the intact and amputated limbs is required to control placement of the amputated leg to achieve a gait comparable to healthy subjects.
Conclusion: There are many factors such as the hip, knee flexion/extension and the ankle plantarflexion/dorsiflexion contributing to the control of the transient impact of an amputee during walking. Therefore, for enhanced control of a prosthetic knee, a multifaceted approach is required. This study showed that UTFA adaption to different prosthetic knees in the short term with slower than self-selected speed is completely achievable based on the negative correlation of ground reaction forces versus linear velocity. Reduced speed may provide the prosthetists with the vision of the amputees' progression of adaptation with a newly prescribed prosthetic knee.

1. Introduction

After amputation and a period of recovery and rehabilitation, a unilateral transfemoral amputee (TFA) must adapt to a new prosthetic leg. Prosthetists prescribe a prosthetic leg, which consists of a knee, ankle, and foot, based on their own experiences and manufacturer's specifications. There is no consensus as to which prostheses may best suit an active amputee, either in the short or long term. An amputee's adaptation is a process of becoming familiar with new circumstances and depends on many features including the subject's psychological state [1] as well as the design of the prosthesis [2]. In general, the two most common types of prosthetic knees available on the market are either mechanically passive or actively controlled variable damping (often known as microprocessor-controlled) prostheses [3,4]. There are contradictory results presented in the literature regarding whether a microprocessor knee prosthesis improves a unilateral transfemoral amputees' gait or not. Kaufman et al. [5] have suggested that the microprocessor knee improves the gait and balance of amputees. However, the data reported by Segal et al. [6] has shown insignificant differences between amputee's gait using mechanically

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Table 1
Grouping of the Prosthetic knee and foot used in this study.

Item	Type of prosthetic knee	Type of prosthetic foot and ankle	Component weight (kg)	General Description
A	Ottobock 3R60	Venture College park	0.890/0.585	Mechanically passive polycentric hydraulic knee
B	Ottobock 3R80	Venture College park	1.240/0.585	Single axis passive controlled hydraulic damping
C	Ottobock C-Leg (3C88-3)	Venture College park	1.240/0.585	Single axis actively controlled variable damping (microprocessor)
D	Orion2 Endolite	Venture College park	1.35/0.585	Single axis actively controlled variable damping (microprocessor)
E	Rheo3 Össur®	Venture College park	1.36/0.585	Single axis actively controlled variable damping (microprocessor)
F	Plie2	Venture College park	1.24/0.585	Single axis actively controlled variable damping (microprocessor)

passive or microprocessor-controlled prosthetic knees. Research has shown that in many cases for below the knee amputees [7,8] as well as above the knee amputees [6,9], the joint active range of motion and torques are different from able-bodied subjects and there is a significant asymmetry between amputees' legs when compared to control subjects (CS) [10]. However, making amputees walk symmetrically is questioned by Winter and Sienko [7]; they suggested that any human system with neuromuscular deficiency may not perform optimally under symmetric gait condition; rather a new asymmetric condition being pursued based on his/her residual system and mechanics of his prosthesis. The lack of control over the prosthetic knee may encourage the use of the non-prosthetic leg during the stance phase [11]. Reported studies have shown that the transtibial amputees (TTA) load their intact leg more than the amputated leg during human ambulation, resulting in a compensatory mechanism that protects the amputated side residual tissues which are more prone to injury [12]. Such a mechanism may be the cause of low back pain (LBP) and osteoarthritis (OA) in amputees to whom LBP and OA prevalent are higher than intact individuals [13]. Asymmetry combined with the high frequency of impact forces resulting in internal stress [14,15] are some known factors contributing to the causation of low back pain and OA. The transient impact, which is known as loading rate calculated from the slope of the first peak of the ground reaction force, has been used to characterize soft tissue loading which has been shown to be a cause of injury in runners [16]. Studies of the vertical ground reaction forces (vGRF) in slow walking and in distance runners using the heel to toe foot contact pattern have shown that the first impact peak occurs shortly after the heel contact with a quick peak and drop, which is a potential cause of lower extremity injuries [16,17]. Zadpoor et al. [18] used simulation to show that the foot contact velocity has a larger effect on the first peak of vGRF than mass and velocity of the upper body. One of the first studies that looked into heel velocity was reported by Winter [19], who documented the heel vertical and horizontal velocities in healthy young adults and found both to be approximately zero prior to contact. He suggested that such control is necessary to have a gentle landing by the heel on the ground and to achieve that, motor control must coordinate the multi-linked trajectories of the lower extremity segments.

The importance of understanding and evaluation of short-term adaptation of an active transfemoral amputee using direct measurements of loading rate and the heel velocity during the late swing phase to early stance is twofold:

- Firstly, the relation between the heel velocity and vGRF may be used to provide an understanding of the progression of the motor skills of an active amputee during a short-term adaptation period.
- Secondly, the relationship between the vertical heel velocity and the transient impact can provide us with a possible understanding of the control strategies implemented by the amputee on different prosthetic knees.

It was assumed that the loading rate is a function of the vertical heel velocity. The amputee's motor skills progression (i.e. adaptation) with the different prosthetic knees during short term adaptation has been evaluated regardless of the prosthetic knee design. Therefore, the hypothesis of this study was that finding a negative correlation between vertical heel velocity and transient loading rate of either leg in an active amputee in the short term, is an indication of amputee's control and adaptation to the prosthetic knee.

2. Methods

This study encompassed level ground walking trials with three different speeds; self-selected pace, faster and slower than the self-selected speeds during a two weeks' adaptation period of UTFA and group of control subjects (CS). The six commercial prosthetic knees used in this study are shown in Table 1.

An UTFA (age: 52 years old; height: 1.66 m; weight: 66.7 kg), without any neurological or orthopedic disorder except for his amputation was the participant in this study. His amputation was performed in April of 2009 and it was due to a trauma to the left knee. The amputee patient wore his pair of comfortable hiking shoes during the motion capture trials and all prosthetic knee alignments were based on the worn shoes. The CS consisted of four healthy male subjects (age: 31.5 ± 7.8 years old; height: 175.1 ± 6.4 cm; weight: 81.3 ± 9.0 kg) without any reported gait abnormalities. The participants were instructed to look straight ahead, with their head erect and their arms at their sides in a comfortable position and move them as they desire during the walk. The participants were informed of the type of study and signed the informed consent approved by the University of Leeds Research Office of Good Practice & Ethics. The motion data was collected using 3D motion capture system Qualisys ProReflex MCU240 and Track Manager (QTM) (Gothenburg, Sweden) with 13 Cameras and 2 AMTI (Watertown, MA, USA) force platforms. The motion and force data were collected at 450 Hz and 1200 Hz, respectively. The reflective surface markers were bilaterally placed on both left and right of the anterior superior iliac spine (ASIS) and posterior superior iliac spines (PSIS), sacroiliac joint, femoral condyle, malleolus, the 1st and 5th metatarsal and calcaneus (Fig. 1).

CODA pelvis in Visual3D^{TM} (Germantown, MD, USA) was used to determine the center of the hip joints and the model was constructed based on [20]. The anatomical landmarks sites were palpated on the amputee by a single individual to reduce possible error. The thirteen cameras were calibrated for the walking region of the participants and recorded the locations of passive reflective markers placed at bony prominences for establishing anatomic coordinate systems for the pelvis, thigh, shank, and foot to capture the relative motion of the linked limbs from inferior to the superior. The joint angles and angular velocities of lower extremities were measured and synchronized that the heal contact for visual comparison and understanding of the overall progression of the amputee during short period adaptation. Each measurement was repeated at least 5 times. The heel trajectories were tracked with a marker placed at the superior posterior site of the calcaneus, typically the very back of the heel about 30 mm off the ground. Heel position data in the anterior/posterior direction, i.e. the direction of motion, and vertical direction (Z) were captured for analysis. The magnitudes of ground reaction forces were divided by the

Fig. 1. Marker placement on the amputee.

individual's body weight (normalized) to allow for a fair comparison between participants. All motion data was filtered using a zero phase second-order Butterworth filter with a cut-off frequency of 8 Hz. The raw GRF and kinematics data collected in the laboratory were transferred to MATLAB (R2015b, The Mathworks, MA, USA) to calculate the loading rates, the speed at which forces impact the body, using (Eq. 1) which is the slope of the first peak of the vGRF. A custom program was written to identify the first peak in the vertical GRFs. Once the point was identified, the developed algorithm in the program used at least 4 points prior to the peak to calculate *average loading rate.* The average loading rate was calculated as the slope of the points divided by the corresponding time differences as is shown in below equation [21].

$$\delta f_i = \frac{F_n - F_{n-1}}{t_n - t_{n-1}} \quad (1)$$

where δf_i is the loading rate for every point along the first peak, F_n is the vGRF and i *and* n are the indices correspond to number of samples. The impact and the heel linear velocities were calculated as described by Eq. (1). Data were tested for significant mean differences among the six knees for the analysis of the variance (ANOVA) based on a number of 6 independent trials. Anderson–Darling were run to test the data distribution is normal and the mean differences were comparable. All analyses were performed in MATLAB (R2016b, The Mathworks, MA, USA). A statistical significance level of 5% was used for the analysis. The adaptation was established by plotting transient impact (dependent variable) versus the heel velocity (independent variable). A negative trend was considered an indication of adaptation as such trend shows vGRF and vertical velocity both decreased, simultaneously.

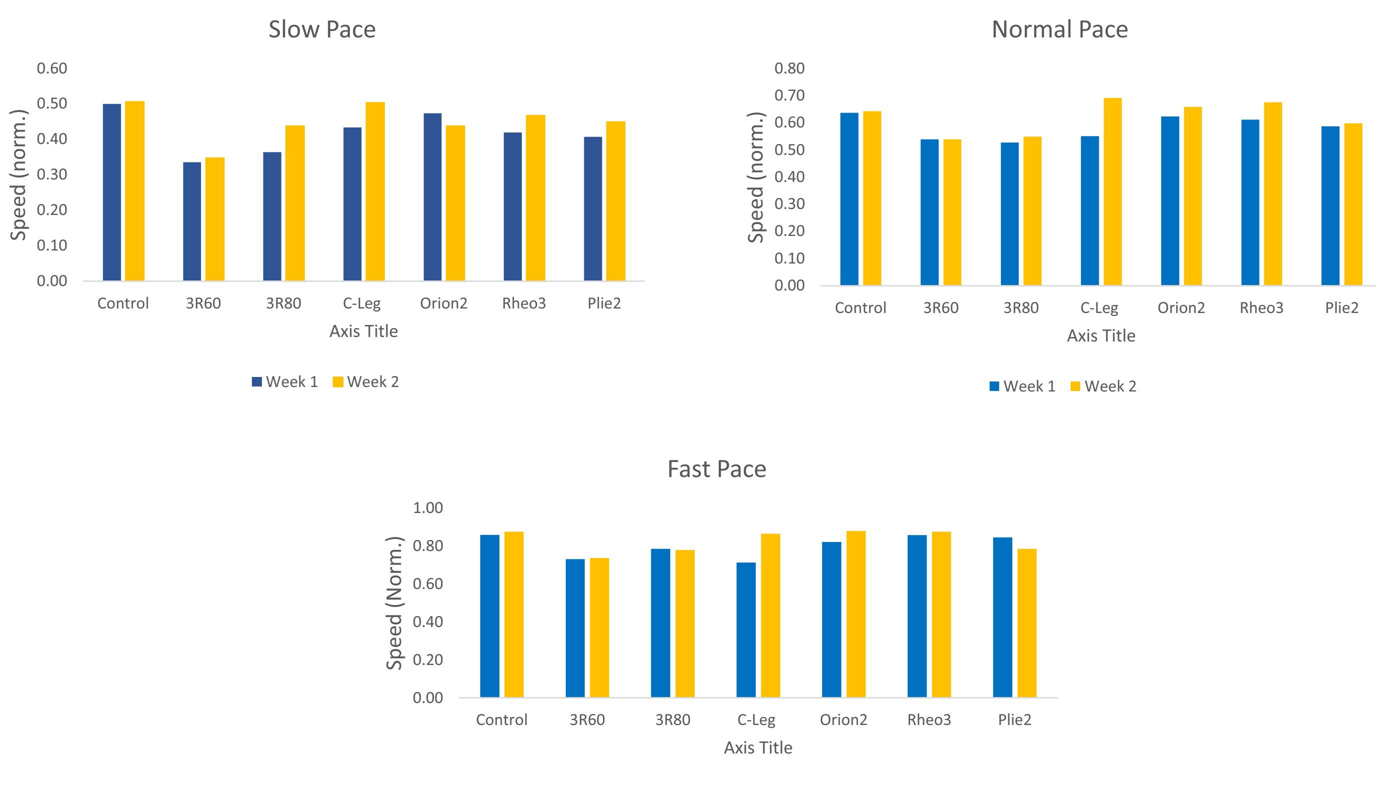

Fig. 2. Amputee's speed with different commercial prosthetic knee versus control subjects (CS).

3. Results

The results are divided into those taken the day the amputee had been fitted with the new prosthetic knee and those taken two weeks later. The gait speed was normalized by the leg length which was measured from the coordinate of the hip joint to the foot heel, given in Fig. 2.

The amputee's speed with microprocessor knees showed an increase after two weeks of adaptation except for the Plie (during the faster pace) and Orion2 (during slower pace). The greater speed

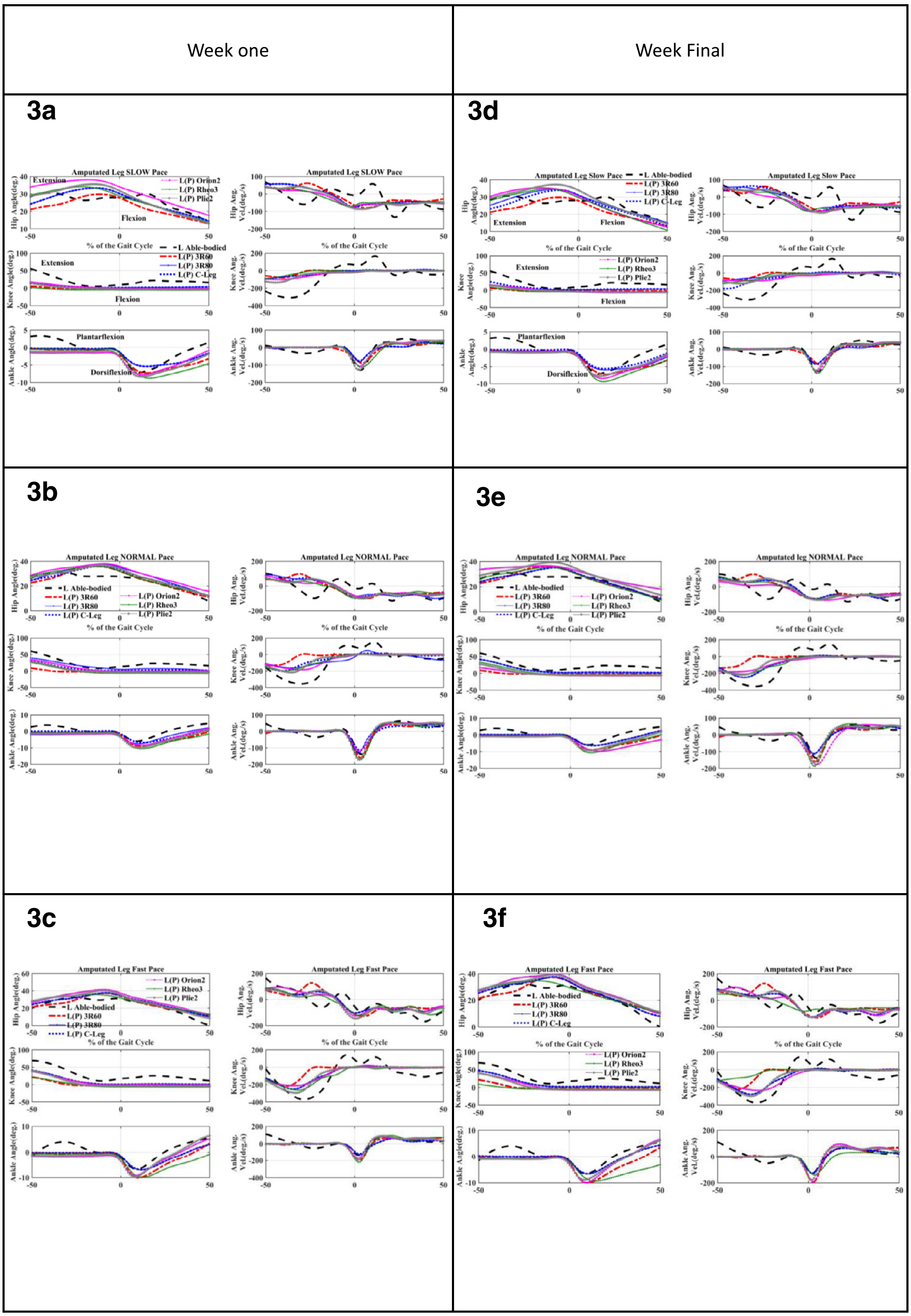

Fig. 3. Range of motion and angular velocity for the hip, knee, and ankle of the *prosthetic side* of three different walking speeds are shown for the amputee with different prosthetic knees and control subjects. The signals were synchronized at the heel contact which is at mid of figures (zero percent). For visualization purposes, only the mean values are shown.

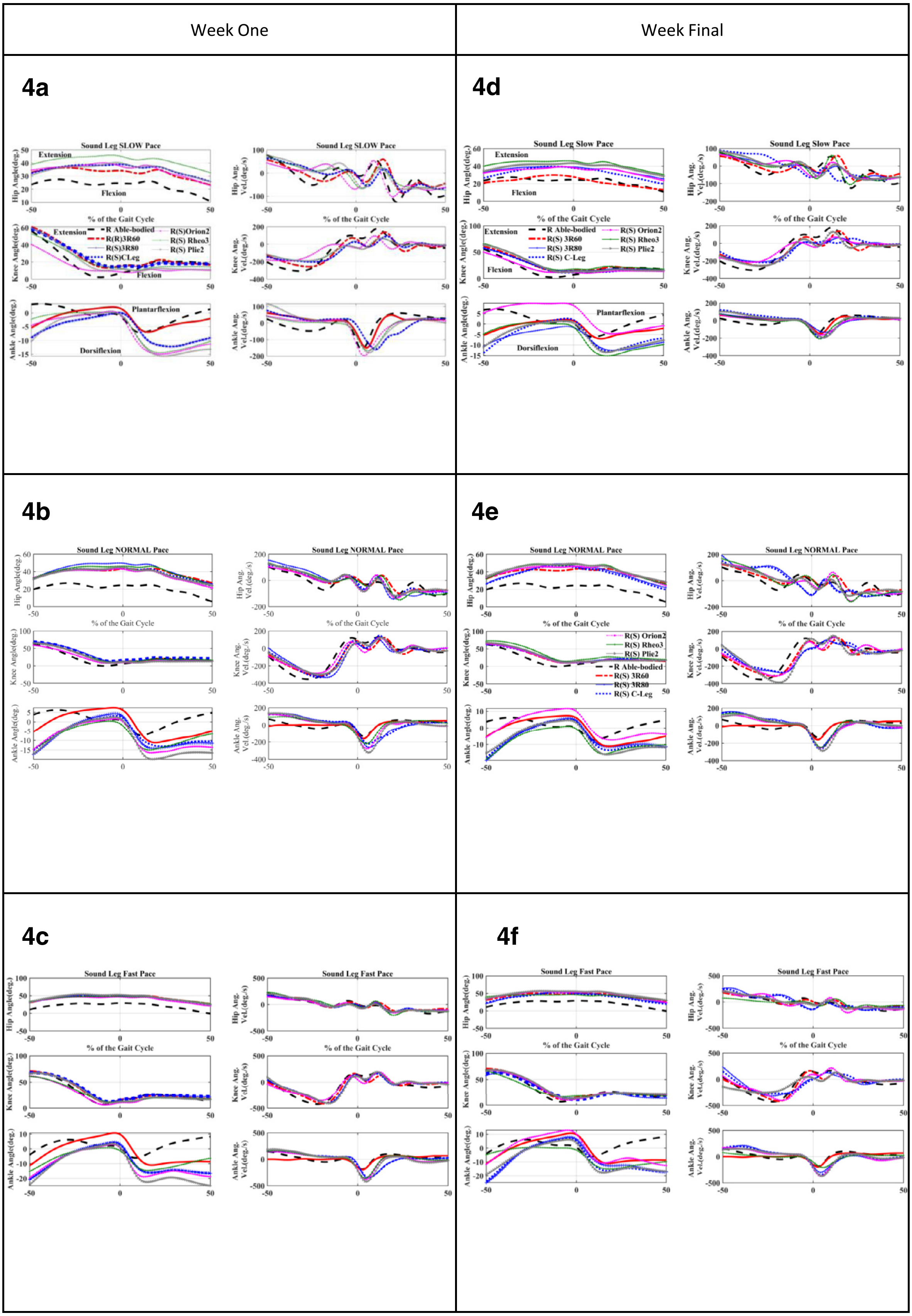

Fig. 4. Range of motion and angular velocity for the hip, knee, and ankle of the *sound side* of three different walking speeds are shown for the amputee with different prosthetic knees and control subjects. The signals were synchronized at the heel contact which is at mid of figures (zero percent). For visualization purposes, only the mean values are shown.

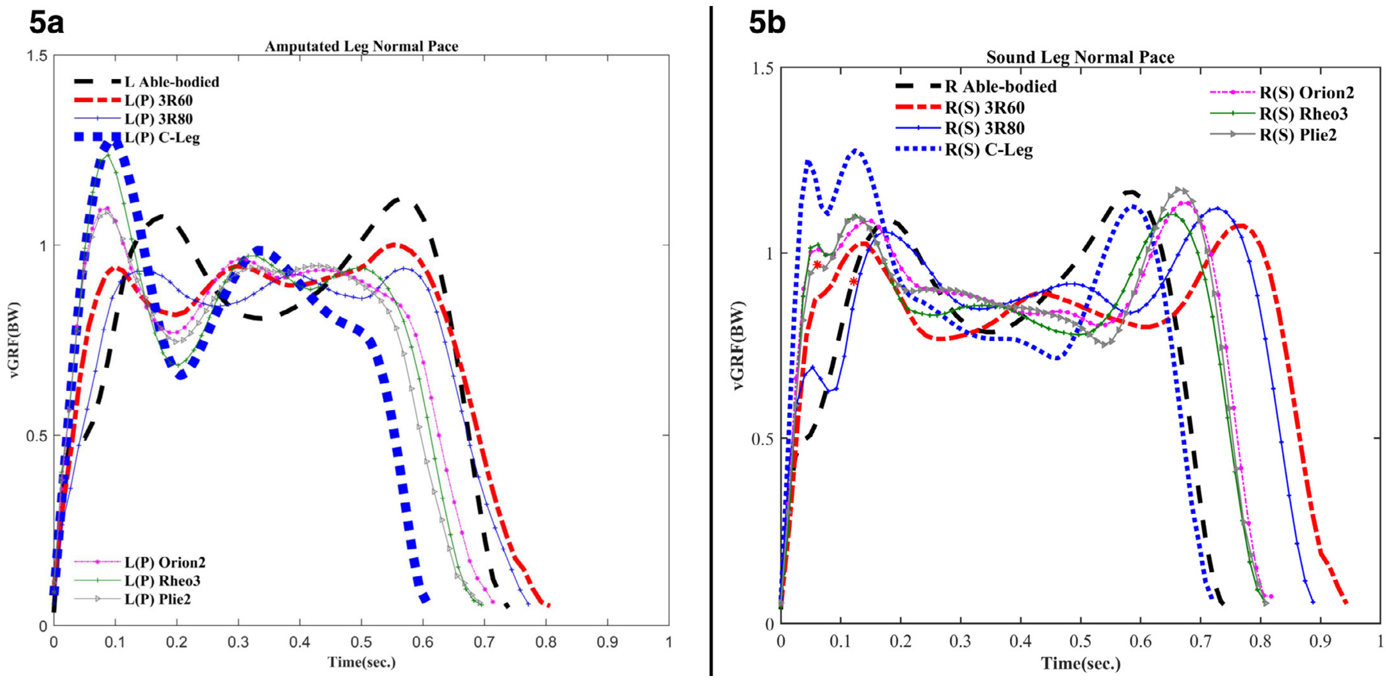

Fig. 5. Mean vGRF normalized to the weight of the subjects versus time for different prosthetic knees and the control subjects during normal pace walking. The loading rate was calculated from the first peak of the plot right after the heel contact has taken place.

values after the adaptation period reflect a greater control over the prosthetic knee.

To obtain a proper comparison between the collected signals, as all analysis was focused before and during the heel contact; all signals were synchronized at the heel contact event (zero percent of the gait cycle). Figs. 3 and 4 depict joint angles range of motion (ROM) and angular velocity obtained during different walking speeds along with control subjects.

The hip, knee, and ankle joint angles showed that there was a similar flexion/extension pattern on both legs. However, the knee on the amputated side showed very little flexion during the swing phase and almost no flexion after the heel contact (20/5° vs 60/20°). The ankle on the amputated side between −50% and −15% of the gait cycle had no activity, contrary to the CS ankles. The trail hip range of motion (ROM) increased about −40% from 30 to almost 39° just before the heel contact while the CS' hips continued to extend to the −20-degree angle. The hip, knee, and ankle joint angular velocities on the amputated leg were less active when compared to the CS.

The results of mean vGRF versus time for the amputee with different prosthetic knees, and the CS when walking with self-selected pace are shown (Fig. 5). The first peak for both prosthetic and the intact leg were larger than the CS.

Fig. 6 shows the box and Whisker plot of the loading rate for different speeds and prosthetic knees.

The loading rates on the intact side, on average, were higher in magnitude in week one for all different speeds. However, in the final week, the contralateral and ipsilateral showed reduced asymmetry, except for the C-Leg for which the intact side produced the mean magnitude of the loading rate twice as large as the prosthetic leg. It appeared that the amputee learned to rely on the intact leg during normal and faster pace. During the slow and normal gait speeds, the mean differences on the sound leg were not significantly different from the CS. The heel vertical displacement, and anterior/posterior (A/P) velocity for the amputee along CS are shown in Fig. 7.

As the gait speed increased, the mean value was significantly different from CS on the amputated side based on our ANOVA study and this was more pronounced in the microprocessor knees. Fig. 8 shows the mean changes in normalized walking velocity for all prosthetic knees against the transient impact.

The plots were connected to show adaptation period history. The fitted lines in both left and right legs were consistent in showing negative correlations between the dependent and independent variables except for the prosthetic leg during the normal speed, and the intact leg during the fast pace. It can be observed that transient impact is not a linear function of the heel velocity.

4. Discussion

This case study aimed to investigate the effect of different commercial prosthetic knees on the short-term progression of a UTFA based on biomechanical parameters. There are many commercial prosthetic knees available today and it is difficult to draw any conclusions on how they may influence the functionality and performance of an active amputee. Our major aim was to monitor the relationship between the vertical heel velocity and the transient impact to draw a conclusion about the adaptation and progression of an UTFA. This is the first study to measure the heel vertical and horizontal trajectories of an amputee and a group of healthy individuals since Winter's study [19] which was conducted for a few healthy participants. Moreover, it was concluded that there are other factors such as the hip and knee flexion/extension and the ankle plantarflexion/dorsiflexion that contribute to the control of the loading rate. It was obvious that motor control played an important role in coordinating the joint just before and during the heel contact to produce a controllable transient impact as the amputee became adapted to the prosthetic knees and learned to slow down his feet [19].

Most of the past studies have focused on a pool of amputees with different types of prosthetic knee and foot with which the biomechanical responses were measured. However, to our

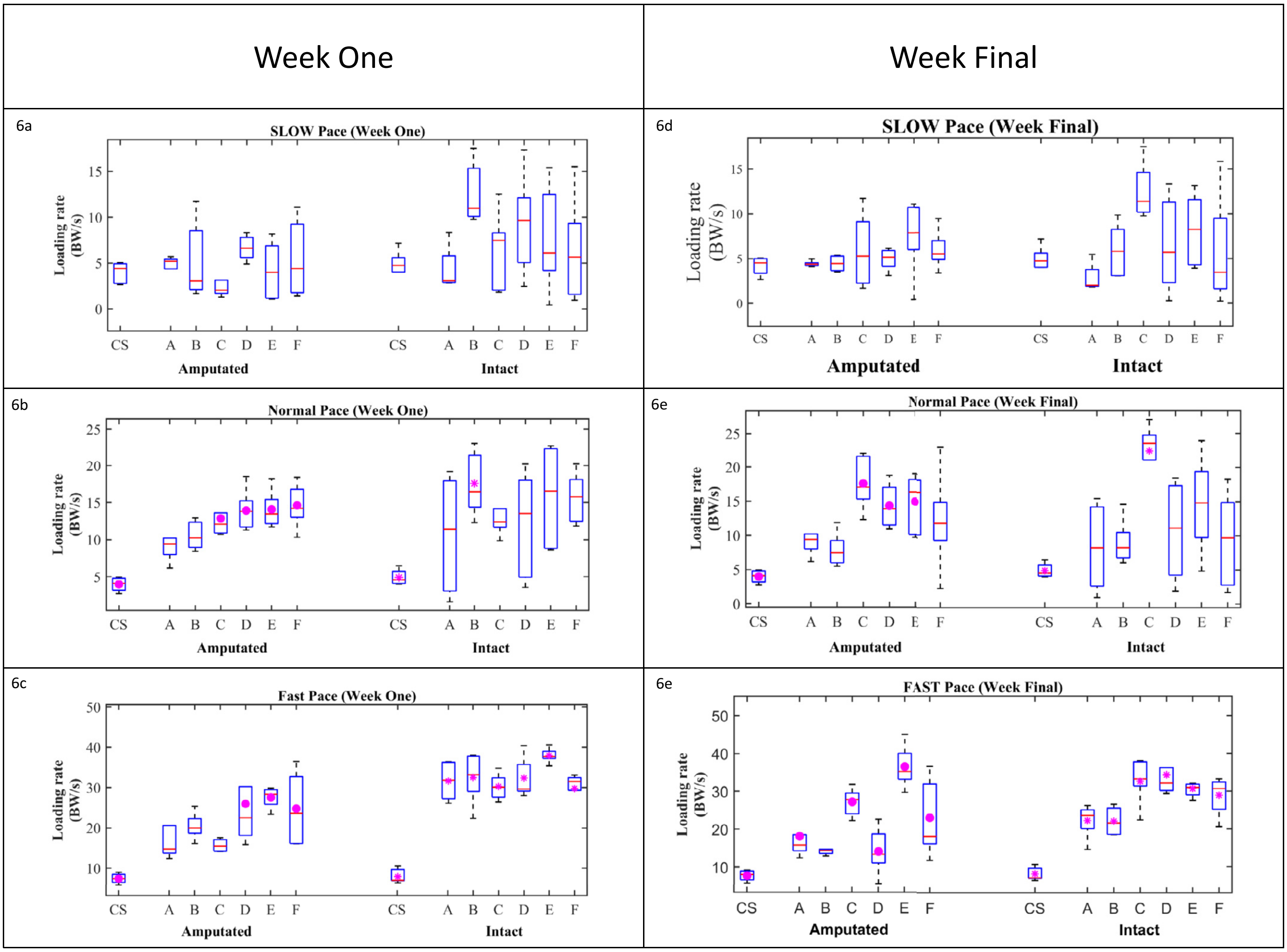

Fig. 6. Left and right loading rates measured for different pace for the amputee with different prosthetic knees versus control subjects. CS: control subjects, an amputee with A: 3R60 B: 3R80, C: C-Leg D: Orion2 E: Rheo3 G: Plie2 knees. The purple circle and star are an indication of mean significant differences between the control subjects and the amputee. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

knowledge, no study has considered how an active TFA may adapt to different prosthetic knees based on direct biomechanical responses during a short period of rehabilitation.

Despite asymmetry in ipsilateral and contralateral legs, the biomechanical adaptation was evident after a short period. The increase in walking speed is usually achieved by larger plantarflexor angle during the push-off. Similar mobility may be achieved by extending the hip to create a flexor torque about the hip joint [19] to prepare the leg for lack of plantarflexion during the push-off, which appears to be the amputee's strategy.

The trend line between the loading rate and the heel's vertical velocity were negatively correlated (Fig. 8) during slow walking showing adaptation of amputee to the prosthetic knees. This may suggest an appropriate rehabilitation strategy to avoid any undesirable adaptation. By prescribing a slower than self-selected speed for the amputee at the beginning of a new prosthetic knee (as she/he becomes more familiar with it) based on their biomechanical responses, the prosthetist may allow the amputee to walk with a faster speed. The loading rate was significantly higher than CS during the normal and faster pace, the values were even more pronounced in the amputee with a microprocessor-based prosthetic knee. By the final week of the adaptation period, the mean loading rates reduced between the amputated and sound legs. With the self-selected and faster speeds, the mean loading rates on the intact side were higher in the first week of adaptation than the prosthetic side. However, this trend changed except for the C-leg during the normal pace, returned to larger loading on the intact leg similar to that reported by the authors in [13,23] except for Rheo2. The Amputees lack of proprioceptive feedback during the gait initiation is also another causation to generate large loading rates [24] in addition to the design and alignment challenges [12]. A study by Keller [25] has shown the group of joggers whose speed were between 1.5 and 6.0 m/s generated loading rates among 8 to 30 BW. The amputee produced larger loading rates comparable to those joggers. It is believed that generating such a large loading rate in UTFA may be a possible cause of injuries to the amputee's musculoskeletal system [11,26].

The amputee's hip on the amputated side has shown a relatively larger ROM in all walking speeds when compared to the CS, which may translate to larger metabolic energy consumption. It appears that, contrary to the data reported by Winter, the vertical heel velocity is not virtually zero for either the CS [19] or the amputee with different prosthetic knees just before the heel contact. Based on Figs. 3 and 4, the amputee's hip on the prosthetic side

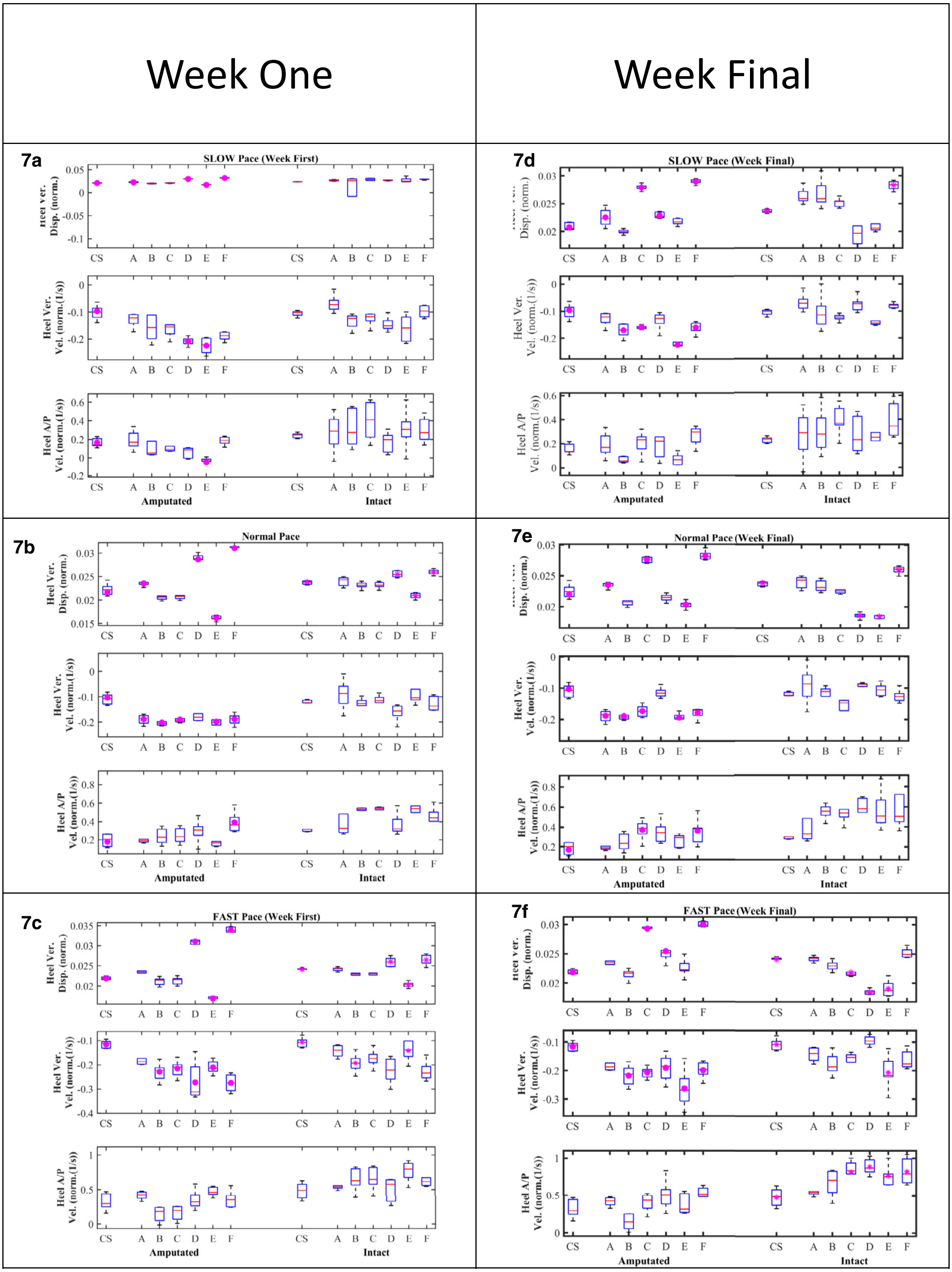

Fig. 7. Heel displacement, vertical velocity and anterior/posterior velocity of the amputee with different prosthetic knees during the adaptation period. CS: Control Subjects A: 3R60 B: 3R80, C: C-Leg D: Orion E: Rheo G: Plie prosthetic knees. The purple circle and star are an indication of mean significant differences between the control subjects and the amputee. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

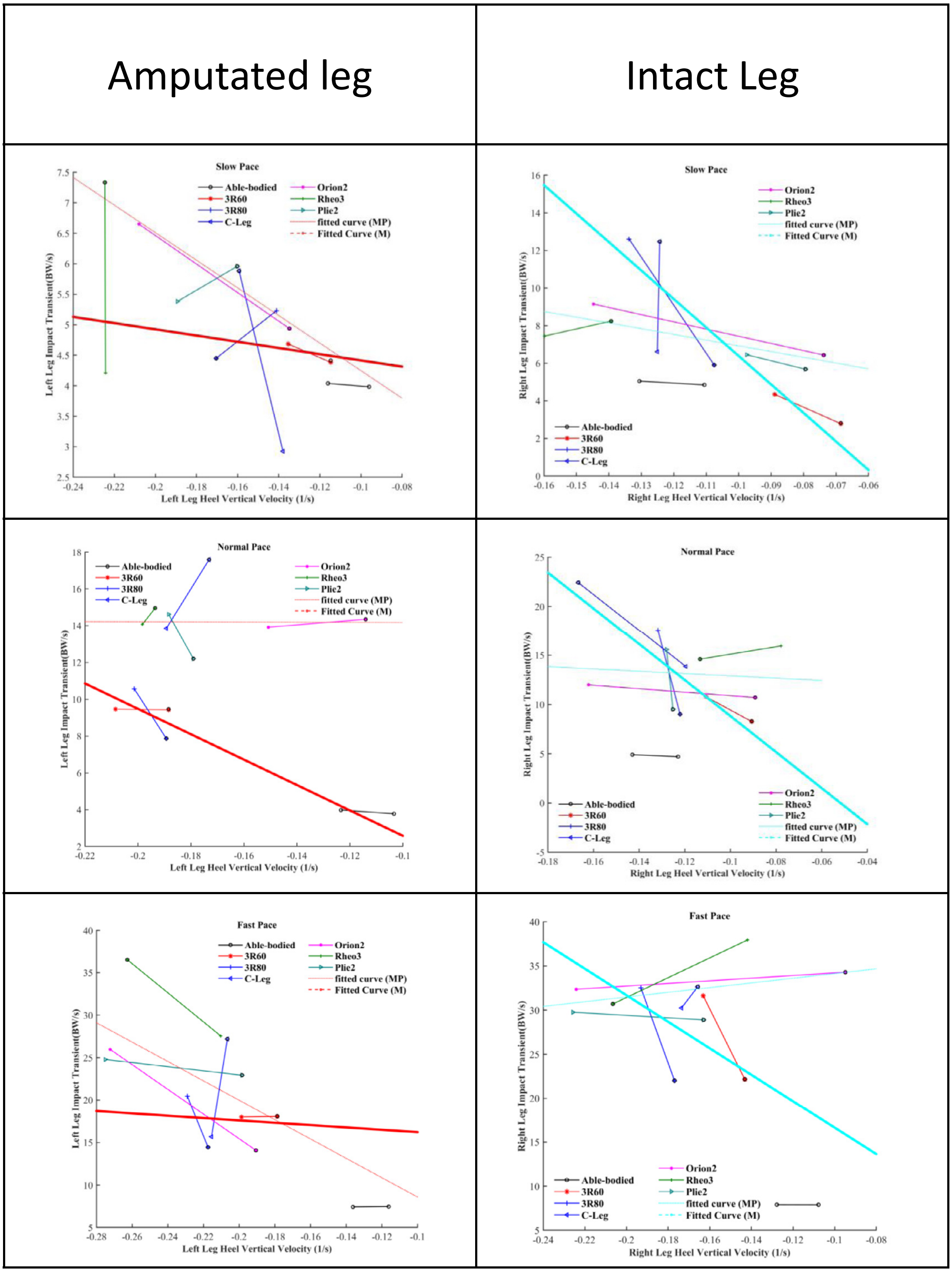

Fig. 8. Relationship between the impact transient and heel vertical velocity. Individuals regression lines were fitted for MP: Microprocessor Knees M: Mechanical passive knees. The final week adaptation results were marked by a black circle around corresponding markers. Both impact transient and heel vertical velocity were normalized to the weight of the participants and the leg length, respectively. Amputee showed remarkable adaptation for both the amputated and the intact limb by the negative correlations observed, except for normal pace in amputated side and the fast pace on the intact side which the correlation for MP knees is positive. The positive correlation showed that despite reduction in transient impact loading rate, the heel vertical velocity did not reduce.

went to an early extension and continued about 10% of the gait cycle after the heel contact occurred (Fig. 3). The foot was relatively stationary in A/P velocity (Fig. 7), however, the highest speed in the late swing phase suggested that the hip reached its greatest speed at −9% before the foot contact. This phenomenon may be explained by dynamic walking in which the lack of plantarflexion on the amputated side is compensated by the hip torque generated against the stance leg [22]. An inspection of the amputee's hip angular rotation with different prosthetic knees indicated that the amputee's hip continued to extend 14% of the gait cycle before the heel contact, which is contrary to CS where the hip was momentarily stationary just before the heel contact occurred (Fig. 3). It appears that the mechanism for a gentle contact of the heel requires significant control and coordination between the knee flexion and ankle plantarflexion [7]. However, this described mechanism was not achieved by the amputee. It is conclusive that excellent motor control is required for the amputee to avoid shock as they are unable to slow down the hip angular velocity to a stationary position and reduce the heel vertical velocity during the late swing phase. The current findings may also suggest why amputees have a tendency to rely on their intact leg more than the amputated leg [27]. The current data appears to support Winter's third motor functions fact, indicating that in order to have a safe control of foot trajectories and to achieve a gentle heel landing, significant motor functions are required to coordinate the hip, the knee, and ankle. The inclusion of a haptic system in the prosthetic knee may enhance amputee's proprioception which may help to reduce the heel velocity. The A/P heel velocity just before the heel contact on average was higher than CS on the intact limb. Elderly postural stability is known to shift from the ankle to the hip due to loss of proprioceptive, and cutaneous inputs [28] resulting in larger A/P heel velocity compared to fit adults [29]. During late swing, activation of the hamstrings group causes a flexion moment at the knee, and an extension moment at the hip, both of which may contribute to the reduction of the anterior/posterior (A/P) velocity of the foot prior to heel contact (HC) [30,31] in amputees. The horizontal heel velocity was reduced slightly in the amputee by the final week of adaptation showing the progression of the amputee to adapt to the prosthetic knees maintaining postural stability.

For our study to generalize the outcome, the number of TFA must increase. However, the approach we chose was to observe and implement a method for how an active amputee's gait will adapt to the prosthetic knees without introducing confounding parameters that may vary from one amputee to another due to the length of amputation, method, and level of confidence during walking. Our focus was given to objective functions and observable parameters that are captured from a motion capture system. This perhaps can be told as the "design of experiment" on the prosthetic knee system. The biomechanical responses showed that the amputee learned to adapt to the prosthetic knees. The amputee adaptation time in this study was only two weeks, however, this has been shown in a study by Barnett et al. [32] (for non-microprocessor devices) to be an adequate time to become adapted to a prosthetic device. The prosthetic alignment is another challenging task which requires to be implemented by an expert prosthetist. To be consistent, all alignments were done in the same hospital with a designated specialist to avoid any possible misalignment. In addition to the above limitation, Uncertainty in placing the passive markers on anatomical sites may be considered another limiting factor, this effect was reduced by considering only one individual to perform such tasks. To estimate kinematics of human joints, motion capture systems currently rely on built-in biomechanical models which consider the joints to be idealized mechanical joint (i.e. revolute and cylindrical joints) rather than a physiological joint performing simultaneous translation and rotation, resulting in reduced kinematics accuracy than what is anticipated. However, to avoid inconsistency, the biomechanical model was considered to be the same for the models and the definition of passive markers was only used to describe the height, mediolateral width of the limb and center of rotation of the joints regardless of the mechanical property of the prosthetic legs or muscle masses on the results.

Conflict of interest

The authors declare no conflicts of interests. The founding sponsors had no role in the design of the study; in the collection, analyses, or interpretation of data, and in the decision to publish the results.

Ethical approval

MEEC-14-011 Faculty Research Ethics Committee at the University of Leeds.

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