

## Comparison of gait of persons with partial foot amputation wearing prosthesis to matched control group: Observational study

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**Abstract**—Our understanding of the gait mechanics of persons with partial foot amputation and the influence of prosthetic intervention has been limited by the reporting of isolated gait parameters in specific amputation levels and limited interpretation and discussion of results. This observational study aimed to more completely describe the gait patterns of persons with partial foot amputation wearing their existing prosthesis and footwear in comparison with a nonamputee control group. Major adaptations occurred once the metatarsal heads were compromised. Persons with transmetatarsal and Lisfranc amputation who were wearing insoles and slipper sockets maintained the center of pressure behind the end of the residuum until after contralateral heel contact. This gait pattern may be a useful adaptation to protect the residuum, moderate the requirement of the calf musculature, or compensate for the compliance of the forefoot. Power generation across the affected ankle was virtually negligible, necessitating increased power generation across the hip joints. The clamshell devices fitted to the persons with Chopart amputation restored their effective foot length and normalized many aspects of gait. These persons' ability to adopt this gait pattern may be the result of the broad anterior shell of the socket, a relatively stiff forefoot, and immobilization of the ankle. The hip joints still contributed significantly to the power generation required to walk.

**Key words:** biomechanics, Chopart, gait, Lisfranc, metatarsophalangeal, orthosis, partial feet, partial foot, prosthesis, rehabilitation, transmetatarsal.

### INTRODUCTION

Partial foot amputation (PFA) describes the loss of part of either the fore- or hindfoot [1] and is typically the result of vascular insufficiency secondary to diabetes [2–5]. However, trauma, frostbite, and congenital anomalies are examples of the many other causes of amputation reported in the literature [6–8].

PFA is the most common amputation in Australia, eclipsing the combined incidence of transtibial and transfemoral amputations [9]. In the 2004 to 2005 calendar year, in excess of 5,300 partial foot procedures were performed, the vast majority of which were amputations of the toe (63.0%) or toe including metatarsal (30.0%), with more proximal amputations, such as transmetatarsal (5.3%) or midtarsal (1.6%), less commonly performed [9]. Based on statistics from the Australian Institute of Health and Welfare (2004–2005), the rate of PFA begins increasing dramatically after about age 40, almost in parallel with the increasing incidence of diabetes.

**Abbreviations:** CI = confidence interval, CoP = center of pressure, GC = gait cycle, GRF = ground reaction force, ML = medial-lateral, MTP = metatarsophalangeal, PFA = partial foot amputation, QALS = Queensland Amputee Limb Service, SD = standard deviation, TMT = transmetatarsal, WV = walking velocity.

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Despite the prevalence of PFA, our understanding of the resulting gait pattern and the influence of prosthetic and orthotic intervention is very much in its infancy [10]. For most of the last few decades, what was known about PFA gait was based on theoretical analyses that stemmed from an appreciation of nondisabled gait [11–13].

The current body of literature consists predominantly of observational studies [10], marking a significant shift toward improvement of the evidence base. However, a number of factors limit the confidence in the outcomes of these investigations and the comprehensiveness of the available evidence.

The confidence able to be invested in the outcomes of these investigations has been compromised by the small and heterogeneous nature of the population with PFA and a number of consistent methodological issues. While these issues have been described in detail as part of a recent systematic literature review [10], some illustrative examples include poor matching of the control and amputee cohorts to account for the effect of systemic diseases such as diabetes and peripheral vascular disease and the inclusion of bilateral amputees in a predominantly unilateral amputee cohort without demonstrating the reasonableness of pooling data in this way.

The comprehensiveness of the evidence base is limited because investigators have tended to describe isolated aspects of gait dealing with specific hypotheses. More comprehensive gait analyses including kinematic and kinetic patterns at the knee and hip joints [14] or the contralateral lower limb [15] are scarce and poorly understood. As such, the purpose of this observational study was to more completely describe the gait patterns of a broad cohort of partial foot amputees with a view to better understanding the underlying mechanical adaptations to PFA and prosthetic fitting.

## METHODS

### Subjects

Subjects with amputation were recruited through either the Queensland Amputee Limb Service (QALS) or prosthetic/orthotic service providers in Queensland, Australia. Of the 56 individuals identified through these avenues, data could be collected on 7 persons with PFA. Minimal exclusion criteria were applied to the sample with amputation because of the limited number of potential participants; as such, the amputee cohort was quite variable in cause of amputation, amputation level, years

since amputation, number of limbs affected, and types of prosthetic fitting (**Table 1**). Subjects were excluded if they ambulated with the use of any gait aids, had concomitant health problems such as ulceration, or had neuromuscular/musculoskeletal conditions that might affect their gait. Diabetes or peripheral vascular disease were not considered criteria for exclusion, although none of the subjects with amputation had these systemic illnesses. The reported incidences of gangrene could be traced back to nonvascular causes, such as frostbite or burns.

Because of the limited and variable amputee sample and because previous investigations have not demonstrated the reasonableness of pooling data from individuals with disparate amputation levels and prosthetic interventions, we used a case-series approach to consider the gait of each participant in isolation relative to a non-amputee control sample.

The control subjects were excluded if, like the persons with amputation, they ambulated with the use of any gait aids, had concomitant health problems such as ulceration, or had neuromuscular/musculoskeletal conditions that might affect their gait. The control subjects were recruited to ensure a reasonable match in mean age, mass, stature, and sex. Subjects in the control group did not have diabetes or vascular disease, and as such, the subjects with amputation and the control subjects were reasonably matched. The mean  $\pm$  standard deviation (SD) age, stature, and mass of the control sample were  $41.13 \pm 14.81$  years,  $1.74 \pm 0.08$  m, and  $77.11 \pm 6.83$  kg, respectively.

### Apparatus

Kinematic data were collected using a Peak six-camera three-dimensional motion analysis system and Motus version 4.3.0 software (Peak Performance Technologies; Centennial, Colorado). This camera setup sampled the location of 20 mm retroreflective markers at a rate of 50 Hz. An OR6-5 six-channel strain gauge force platform and amplifier (Advanced Mechanical Technology Inc; Watertown, Massachusetts) were used to sample ground reaction forces (GRFs) at a rate of 1,000 Hz, and data were low-pass filtered at the amplifier with the cutoff frequency of 1,050 Hz inherent to the hardware. The Peak-Motus software controlled the synchronization of the kinematic and externally measured force data.

Custom linked-segment models that accounted for the anthropometry of the residual foot, proximal limb segments, prosthetic fitting, and footwear were developed to enhance the accuracy of kinetic calculations [16].

**Table 1.**

Characteristics of subjects with amputation.

Subject	Level	Years Since Amputation	Etiology	Age (yr)	Stature (m)	Mass (kg)	Device
1004-1307A	MTP*	13	Gangrene <sup>†</sup>	40	1.7	64.9	Full-length shoe insert with PP sole, EVA upper and toe block, silicone pad under metatarsal ends.
2103-2116A	TMT	3	Trauma	54	1.8	84.5	EVA toe filler.
2703-1903A	Lisfranc	22	Trauma	53	1.8	76.6	Acrylic resin below ankle slipper socket with silicone liner. Prosthetic forefoot bonded onto socket.
0704-0403A	Lisfranc	5	Trauma	22	1.8	81.5	Carbon fiber below ankle socket with flexible acrylic resin bootie that extends above ankle. EVA toe filler without reinforcing toe plate.
2103-1906A	Lisfranc	5	Trauma	55	1.8	80.7	Shoe stuffed with sock.
0904-1924A	Chopart*	31	Gangrene <sup>‡</sup>	31	1.7	82.2	Acrylic clamshell PTB with posterior opening. Laminated extension onto socket that extended to toe break. Remainder of forefoot length made from EVA.
3004-1102A	Chopart	12	Trauma	19	1.8	93.0	Acrylic clamshell PTB with posterior window. Sole and forefoot made from SACH foot ground to accommodate socket.
Mean $\pm$ SD	—	13 $\pm$ 10	—	39 $\pm$ 15	1.8 $\pm$ 0.1	80.5 $\pm$ 8.5	—

\*Bilateral amputation at level described.

<sup>†</sup>Gangrene secondary to frostbite.<sup>‡</sup>Gangrene secondary to water burns.

EVA = ethyl vinyl acetate, MTP = metatarsophalangeal, PP = polypropylene, PTB = patellar tendon-bearing, SACH = solid-ankle cushioned heel, SD = standard deviation, TMT = transmetatarsal.

Compared with a conventional linked-segment model, more accurate modeling increased the peak knee flexion and hip extension moment during terminal swing [16]. The magnitude of these differences was comparable to that observed in the gait of nondisabled persons changing from a slow-to-normal or normal-to-fast velocity [17–18].

The anthropometric input data for the partial and sound feet were determined using custom developed anthropometric models [16]. Anthropometric characteristics of the thigh and leg segments were determined using previously published mathematical models based on simple geometric forms [19]. The physical characteristics of the prosthesis and footwear were measured using standard dynamics techniques, including a pendulum trifilar system [20]. Collection of these anthropometric data required anthropometric calipers, a 30 cm ruler, a tape

measure, a plumb line, a stadiometer, electronic scales, and a handheld stopwatch.

Lower-limb joint passive range of motion was evaluated using a goniometer with angles marked in 2° increments.

### Procedures

Participants presented to the biomechanics laboratory and, following explanation of the procedures, provided informed consent as required by the University Human Research Ethics Committee of the Queensland University of Technology.

We interviewed participants to obtain a detailed initial assessment that included details about their amputation, past medical history, and prosthetic management. A qualified prosthetist assessed each participant's residual

foot to determine the amputation level, and where possible, this level was verified through X-ray, surgical reports, or comparison of the residual and sound foot lengths [15]. Details about the type and construction of the prosthesis were also noted (**Table 1**). A qualified prosthetist evaluated the quality of prosthetic fit and function to ensure it was appropriate.

We obtained measures of joint passive range of motion and muscle strength by using standard positions and techniques [21] and making subtle adaptations when forefoot landmarks were absent. As an illustrative example, measures of dorsiflexion and plantar flexion were obtained with one axis of the goniometer aligned parallel to the sole of the residuum rather than through the fifth metatarsal head. Measurements were compared bilaterally and against normative data, as appropriate, to minimize errors [22].

Anthropometric measurements of stature and mass were recorded along with specific foot, leg, and thigh segment measurements as was necessary to provide input data to the anthropometric models. For the leg and thigh models, specific measurements have been described elsewhere [19], but examples include segment length and multiple circumferences and medial-lateral (ML) dimensions along the length of the limb segment. For the sound and partial foot, detailed descriptions of anthropometric measurements have been reported elsewhere [15] but included parameters such as residual foot length, sound foot length, lateral malleolus height above the floor, and ML dimensions across the heel and distal end of the residual foot.

Retroreflective markers were located on the following anatomical landmarks: the spinous process of the fifth lumbar vertebra, the anterior superior iliac spine, the greater trochanter of the femur, the knee joint space inferior to the lateral epicondyle of the femur, the lateral malleolus, the posterior calcaneus at the level of the fifth metatarsal head marker, and the fifth metatarsal head or its estimated location. Markers were also located midhigh and midleg just anterior to the line connecting the proximal and distal segment markers. The location of the absent fifth metatarsal head was duplicated from the sound foot by placement of a ruler posterior to the shod foot and measurement of the distance from the ruler to the center of the marker. In the case of bilateral PFA, the locations of the absent fifth metatarsal heads were estimated with respect to the lateral malleolus by regression equations based on stature [23], which were validated with nonamputees [15]. Markers were located on the outside of the prosthesis over the bony landmarks as was necessary.

The gait of the subjects with amputation was evaluated while they wore their current prosthesis and footwear (**Table 1**). The gait of the control subjects was evaluated while they wore their own footwear. No prostheses were fabricated and no specific footwear was provided for this investigation.

### Data Acquisition and Processing

Participants walked along a level 10 m walkway at their self-selected walking speed until they felt comfortable that they could perform the task. Seven trials were obtained for each lower limb. Trials were repeated if participants adjusted their walking velocity or coordination to strike the force platform.

Three-dimensional marker coordinates were reconstructed and any missing data interpolated with spline routines standard to the Peak-Motus software. Subsequent data processing was undertaken using software written in MATLAB 5.3 (MathWorks Inc; Natick, Massachusetts). Marker displacement data were filtered using a zero-lag, fourth-order Butterworth digital filter with a 6 Hz cutoff frequency [15]. A two-dimensional approach was used, whereby individual segment angles relative to the horizontal were determined with an arc tangent function and joint angles were then determined as the difference between adjacent segment angles [23].

Externally measured force data were filtered using a zero-lag, fourth-order Butterworth digital filter with a 125 Hz cutoff frequency to remove unwanted artifact affecting the signal [15]. Difference in force and moment data from absolute zero were accounted for with offsets determined from a 1-second sample of data collected before initial contact. Data were subsampled to match the sampling frequency of the kinematic data, and as such, further filtering was unnecessary. We used standard techniques to calculate center of pressure (CoP) excursion data when the magnitude of the vertical GRF exceeded 10 N [23].

Custom linked-segment models were used to estimate the kinetics of gait and account for the unique anthropometry of the residual foot, proximal limb segments, prosthesis or orthosis, and footwear. For individuals using below-ankle devices that did not immobilize the ankle, anthropometric descriptions of individual segments (i.e., residual foot and footwear or prosthesis) were combined using standard rigid-body dynamics techniques to create a "lumped" segment that could be described with a single set of body segment parameter data. In this way, the basic assumptions of a standard linked-segment model [23]

remain unaffected. However, for individuals with the clamshell prosthesis, which eliminates ankle motion, the linked-segment model considered the residual foot, leg, prosthesis, and shoe as a single free body hinged at the knee joint under the assumptions that—

- The clamshell prosthesis eliminated ankle motion.
- The prosthesis encompassed the residual foot and the leg segment or a portion of the leg segment.
- The residual limb, leg, prosthesis, and shoe could be considered a lumped free body that rotated around the knee joint and could be described by a single set of anthropometric data such that the mass, center of mass, and mass moment of inertia were combined.
- The center of mass of the lumped free body could be described relative to the knee joint center, and the mass moment of inertia was taken through the center of mass of the lumped segment.

Unfortunately, modeling the clamshell prosthesis in this way eliminates some useful information. For example, reporting ankle moments enables a more complete understanding of the extent to which these devices restore the effective foot length. As such, ankle kinematic and kinetic data have been reported for persons using clamshell prostheses. These data were calculated using a standard linked-segment model that assumed ankle motion [15].

Joint powers were calculated as the scalar product of moment and angular velocity and accounted for power transfer across joints [23]. The resultant components of the joint moments and powers were normalized by body mass [24–25].

The CoP excursion data have been plotted with respect to the gait cycle, rather than the stance phase, to facilitate comparison with the external force and kinetic data. In this way, all data are expressed with respect to the same timescale.

Data obtained for multiple trials were averaged for each limb. The gait of each subject with amputation was considered relative to the mean and 95% confidence interval (CI) of the control cohort.

To avoid overinterpreting and reporting idiosyncratic movement patterns, which can be inherent problems with a case-series approach, we focus the “Results” and “Discussion” sections on the consistent patterns of movement that emerged from studying multiple subjects with amputation. As such, the details associated with the movement patterns of any one subject with amputation have deliberately been overlooked in an attempt to clearly portray the gait of this group as a whole. Where applicable, the movement patterns thought to be reflective of a particular

amputation level or prosthetic intervention have been drawn out in the “Discussion” section.

## RESULTS

The results are presented in discrete data sets, including joint range and muscle strength, temporospatial characteristics, external force, kinematics, and kinetics.

### Joint Range and Muscle Strength

Measured with the Oxford Manual Muscle test, muscle strength on both the sound and affected limbs of the subjects with amputation was typically grade 5—comparable to the control cohort. Reductions in muscle strength were observed in a single subject with Lisfranc amputation: subject 2103-1906A had isolated hip adductor weakness (grade 4).

Meaningful reductions in joint range were isolated to the affected ankle. Plantar flexion range was reduced in the subjects with Chopart amputation. Passive ankle range was limited in the subject with unilateral Chopart amputation (20°) and the subject with bilateral Chopart amputation (mean  $\pm$  SD of both limbs = 30°  $\pm$  3°) compared with the 95% CI of the control group (32°–60°). No significant reductions in dorsiflexion range were observed in the affected limbs of the subjects with amputation compared with the 95% CI of the control group (6°–18°).

### Temporospatial Characteristics

Temporal and spatial descriptors of gait have been presented for the control sample and individual subjects with amputation in **Tables 2** and **3**, respectively.

Reductions in walking velocity, outside the 95% CI of the control sample, were observed in two subjects with amputation, each of a different amputation level (**Table 3**). In the subject with transmetatarsal (TMT) amputation (2103-2116A), reductions in walking velocity (WV) were the result of reductions in stride length not cadence (**Table 3**). In the subject with unilateral Chopart amputation (3004-1102A), reductions in cadence and stride length were not outside the 95% CI of the control group.

In the subjects with amputation, the duration of the gait cycle (GC) remained consistent with that of the control cohort (**Table 2**). Changes in the duration of swing and stance (as a proportion of the GC) as well as single- and double-limb support (as a proportion of the GC) were quite variable between individuals, with no consistent pattern observed (**Table 2**).

**Table 2.**

Temporal characteristics of control subjects and subjects with amputation. Time has been expressed in seconds and in percentage of gait cycle (%GC). Affected limb (AL) and sound limb (SL) values have been reported for subjects with unilateral amputation. In subjects with bilateral amputation, temporal values for each limb have been averaged.

Subject	Level	Gait Cycle (s)	Stance Time		Swing Time		CHC (%GC)	Single Support		Double Support	
			s	%GC	s	%GC		s	%GC	s	%GC
<b>Control</b>		1.176 ± 0.074	0.688 ± 0.048	60.4 ± 0.8	0.451 ± 0.029	39.6 ± 0.8	49.8 ± 0.2	0.452 ± 0.029	39.7 ± 0.8	0.120 ± 0.014	10.5 ± 0.9
<b>Amputation</b>											
1004-1304A	MTP*	1.156 ± 0.005	0.688 ± 0.007	59.6 ± 0.9	0.467 ± 0.012	40.4 ± 0.9	49.7 ± 0.3	0.467 ± 0.013	40.2 ± 1.1	0.114 ± 0.006	9.9 ± 0.6
2103-2116A	TMT										
	AL	1.183	0.720	60.9	0.463	39.1	47.8	0.444	38.0 <sup>†</sup>	0.119	10.2
	SL	1.195	0.750	62.8 <sup>†</sup>	0.445	37.2 <sup>†</sup>	52.8	0.460	39.4	0.156	13.4 <sup>†</sup>
2703-1903A	Lisfranc										
	AL	1.130	0.663	58.7 <sup>†</sup>	0.467	41.3 <sup>†</sup>	49.2	0.452	38.5	0.133	11.3
	SL	1.181	0.731	61.9	0.450	38.1	50.6	0.466	40.8	0.108	9.5
0704-0403A	Lisfranc										
	AL	1.147	0.703	61.3	0.443	38.7	48.9	0.416	36.5 <sup>†</sup>	0.133	11.7
	SL	1.114	0.697	62.6 <sup>†</sup>	0.417	37.4 <sup>†</sup>	50.7	0.443	38.4	0.144	12.4 <sup>†</sup>
2103-1906A	Lisfranc										
	AL	1.105	0.680	61.7	0.423	38.3	51.6	0.443	40.5	0.116	10.6
	SL	1.086	0.644	59.3	0.442	40.7	48.5	0.419	37.8 <sup>†</sup>	0.113	10.2
0904-1924A	Chopart*	1.158 ± 0.032	0.694 ± 0.015	59.9 ± 0.4	0.464 ± 0.017	40.1 ± 0.4	49.3 ± 0.9	0.464 ± 0.015	39.9 ± 0.1	0.123 ± 0.010	10.6 ± 1.2
3004-1102A	Chopart										
	AL	1.228	0.750	61.1	0.478	38.9	50.6	0.461	37.0 <sup>†</sup>	0.165	13.2 <sup>†</sup>
	SL	1.238	0.777	62.7 <sup>†</sup>	0.462	37.3 <sup>†</sup>	49.5	0.478	38.32	0.129	10.3

\*Bilateral amputation at level described.

<sup>†</sup>Outside 95% confidence interval of control group.

CHC = contralateral heel contact, MTP = metatarsophalangeal, TMT = transmetatarsal.

**Table 3.**

Spatial characteristics (mean ± standard deviation) of control subjects and subjects with amputation.

Subject	Level	Cadence (step/min)	Stride Length		Walking Velocity	
			m	Norm	m/s	Norm (s <sup>-1</sup> )
Control	—	105.8 ± 6.9	1.56 ± 0.08	0.90 ± 0.05	1.37 ± 0.07	0.80 ± 0.06
1004-1304A	MTP*	103.9 ± 0.4	1.57 ± 0.02	0.90 ± 0.01	1.36 ± 0.01	0.78 ± 0.01
2103-2116A	TMT	100.9 ± 0.7	1.40 <sup>†</sup> ± 0.05	0.77 <sup>†</sup> ± 0.03	1.18 <sup>†</sup> ± 0.04	0.64 <sup>†</sup> ± 0.02
2703-1903A	Lisfranc	103.9 ± 3.2	1.56 ± 0.08	0.85 ± 0.04	1.35 ± 0.11	0.74 ± 0.06
0704-0403A	Lisfranc	106.2 ± 2.1	1.44 ± 0.02	0.78 ± 0.01	1.28 ± 0.01	0.69 ± 0.01
2103-1906A	Lisfranc	109.7 ± 1.2	1.48 ± 0.01	0.82 ± 0.00	1.35 ± 0.02	0.75 ± 0.01
0904-1924A	Chopart*	103.8 ± 2.9	1.44 ± 0.05	0.83 ± 0.03	1.25 ± 0.08	0.72 ± 0.05
3004-1102A	Chopart	97.34 ± 0.6	1.45 ± 0.05	0.81 ± 0.03	1.18 <sup>†</sup> ± 0.03	0.66 <sup>†</sup> ± 0.02

\*Bilateral amputation at level described.

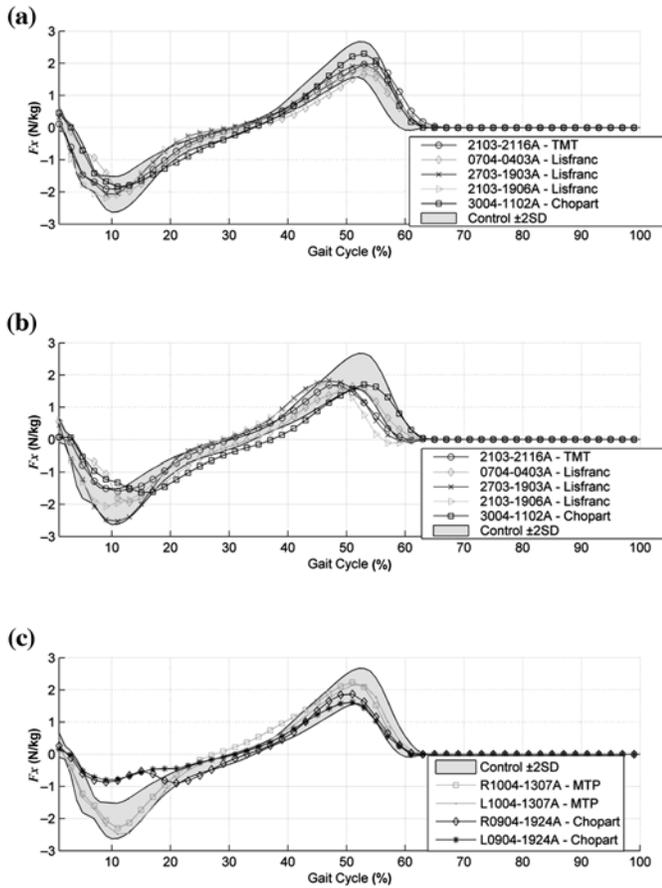
<sup>†</sup>Outside 95% confidence interval of control group.

MTP = metatarsophalangeal, Norm = values normalized by stature in meters, TMT = transmetatarsal.

### Ground Reaction Force and Center of Pressure Excursion Data

Amputation did not influence the horizontal GRF patterns observed in the sound limb (**Figure 1(a)**). The horizontal GRF patterns observed on the affected limb(s) during loading response were quite variable, with timing

of the first peak delayed and the magnitude of the peak reduced more commonly, but not exclusively, in those with bilateral Chopart amputation (**Figure 1(b)–(c)**). During terminal stance, consistent reductions in the magnitude of the horizontal GRF were observed, along with premature timing of the peak, in the subjects with TMT and

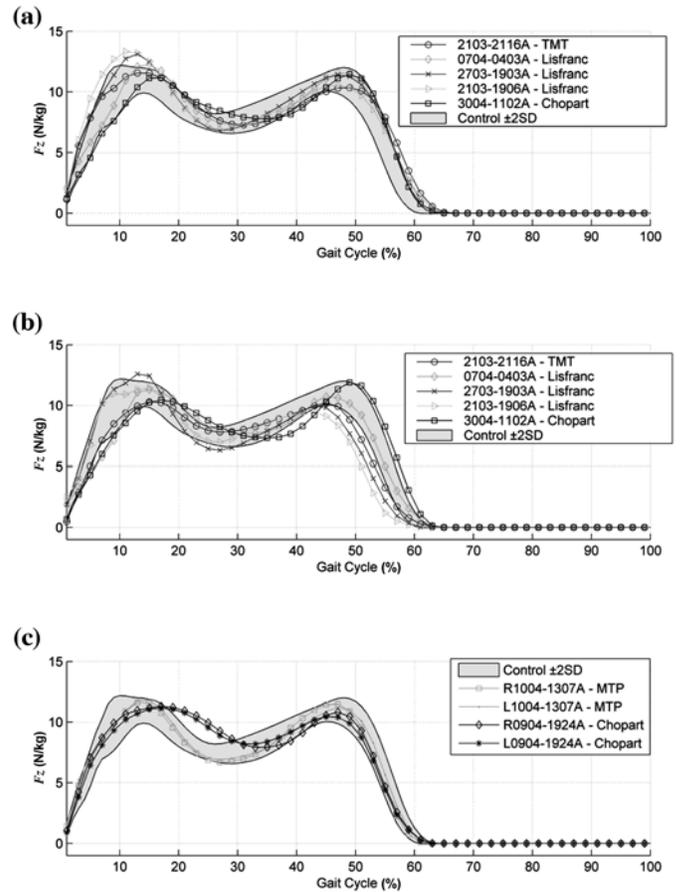


**Figure 1.**

Fore-aft ground reaction force ( $F_x$ ) for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data have been normalized by body mass and expressed in newtons per kilogram. MTP = metatarsophalangeal, TMT = transmetatarsal.

Lisfranc amputation (**Figure 1(b)**). These same characteristics were not evident in the subjects with metatarsophalangeal (MTP) or Chopart amputation (**Figure 1(b)–(c)**).

The vertical GRF patterns observed on the sound limbs were quite variable during loading (**Figure 2(a)**), with the magnitude of the vertical GRF increased in the subjects with Lisfranc amputation beyond the 95% CI of the control cohort. On the affected limbs of the subjects with unilateral amputation, considerable variability was observed between individuals during loading response (**Figure 2(b)**). However during terminal stance, the vertical GRF peak was very consistent and the magnitude was comparable with that observed at the lower end of the control population (**Figure 2b**). The timing of the second



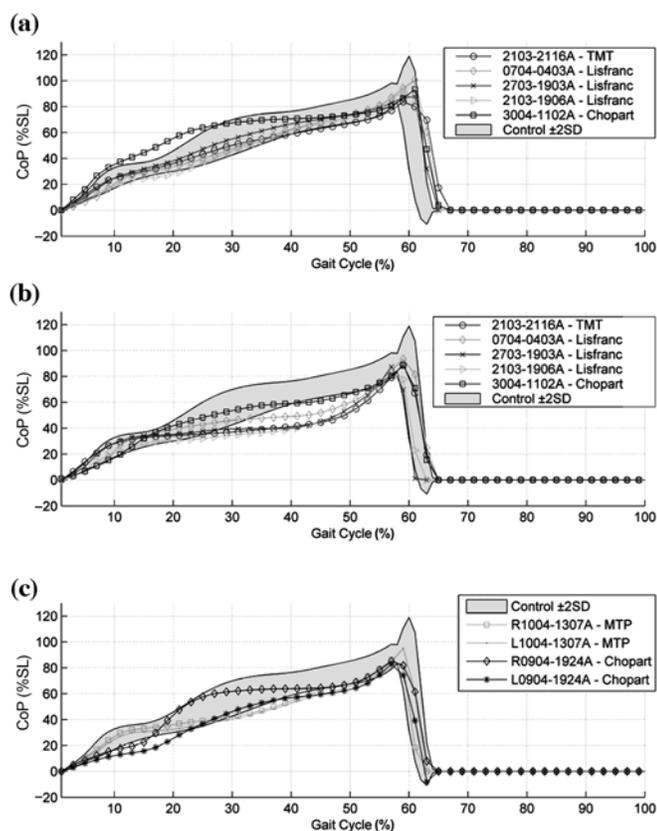
**Figure 2.**

Vertical ground reaction force ( $F_z$ ) for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data have been normalized by body mass and expressed in newtons per kilogram. MTP = metatarsophalangeal, TMT = transmetatarsal. *Source:* **Figure 2(b)** is copyrighted by and reprinted by permission from Informa Healthcare and was originally published in Dillon MP, Barker TM. Can partial foot prostheses effectively restore foot length? *Prosthet Orthot Int.* 2006;30(1):17–23.

peak was premature in all but the Chopart amputee (**Figure 2(b)**). This same pattern was not reflected in the subjects with bilateral amputation of the same amputation level (**Figure 2(c)**).

The CoP excursion patterns on the sound limb were comparable to the control group in all but the subject with Chopart amputation, in whom the GRF force progressed anteriorly along the length of the foot much more rapidly (**Figure 3(a)**). For the affected limbs, the CoP excursion patterns exhibited during loading response

were comparable to that of the control sample, after which more distal progression of the GRF was retarded (**Figure 3(b)**). The degree to which distal progression of the GRF occurred varied with amputation level. In the subject with MTP amputation, the GRF did not progress as distally along the length of the foot, but the normal pattern of CoP excursion was maintained (**Figure 3(c)**). In the subjects with TMT and Lisfranc amputation, the GRF remained at a relatively fixed position (40%–50% of shoe length) until about contralateral heel contact,



**Figure 3.**

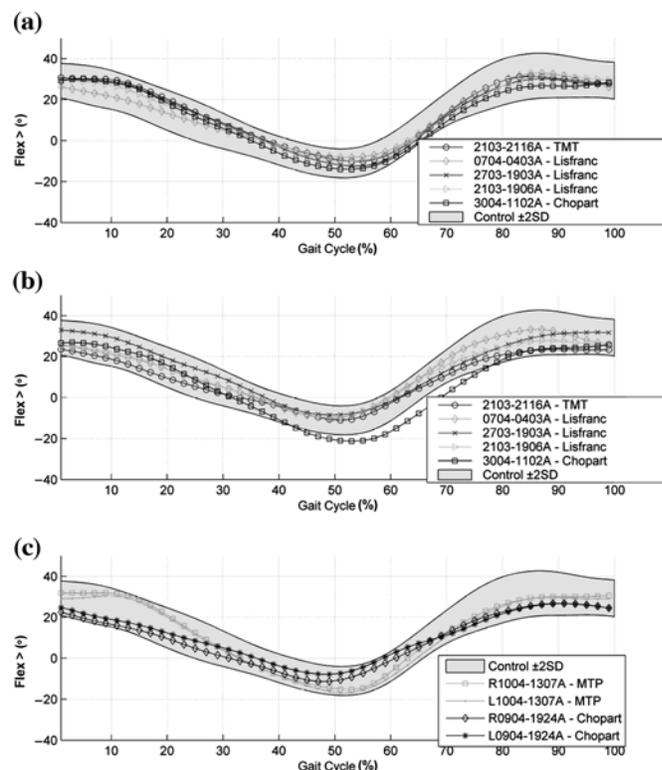
Center of pressure (CoP) excursion for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data normalized by shoe length. CoP excursion data were plotted with respect to gait cycle, rather than stance phase, to facilitate comparison with external force and kinetic data. Note that CoP becomes undefined once foot leaves ground and does not return to heel, which can be interpreted in casual consideration of figure. MTP = metatarsophalangeal, SL = shoe length, TMT = transmetatarsal. Source: Figures are copyrighted by and reprinted by permission from by Informa Healthcare and were originally published in Dillon MP, Barker TM. Can partial foot prostheses effectively restore foot length? Prosthet Orthot Int. 2006;30(1):17–23.

which occurred at 50 percent GC (**Figure 3(b)**). By contrast, the subjects with Chopart amputation exhibited a more normal CoP excursion pattern despite the more proximal amputation (**Figure 3(b)–(c)**).

### Kinematic Data

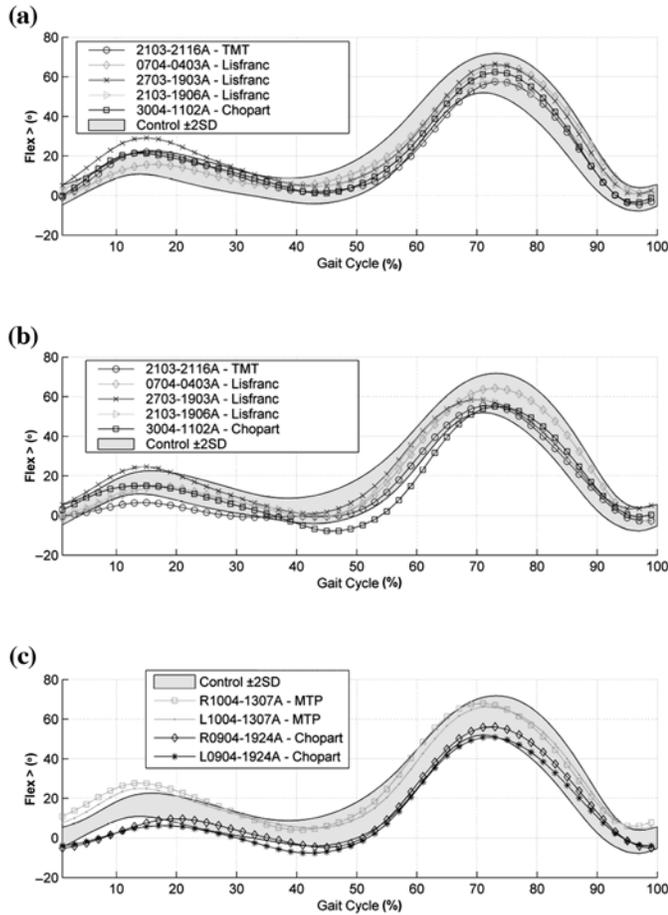
The movement patterns exhibited at the hip and knee joints for the sound and affected limbs were comparable to the control group, aside from a few idiosyncrasies (**Figures 4–5**). Of note was the knee hyperextension observed on the affected limbs of the subjects with Chopart amputation (**Figure 5(b)–(c)**).

The sound-limb ankle kinematic patterns were similar to those observed for the control sample aside from some idiosyncratic movement patterns, such as those exhibited by the subject with unilateral Chopart amputation (**Figure 6(a)**). Peak ankle dorsiflexion was delayed and exaggerated compared with the control group on the



**Figure 4.**

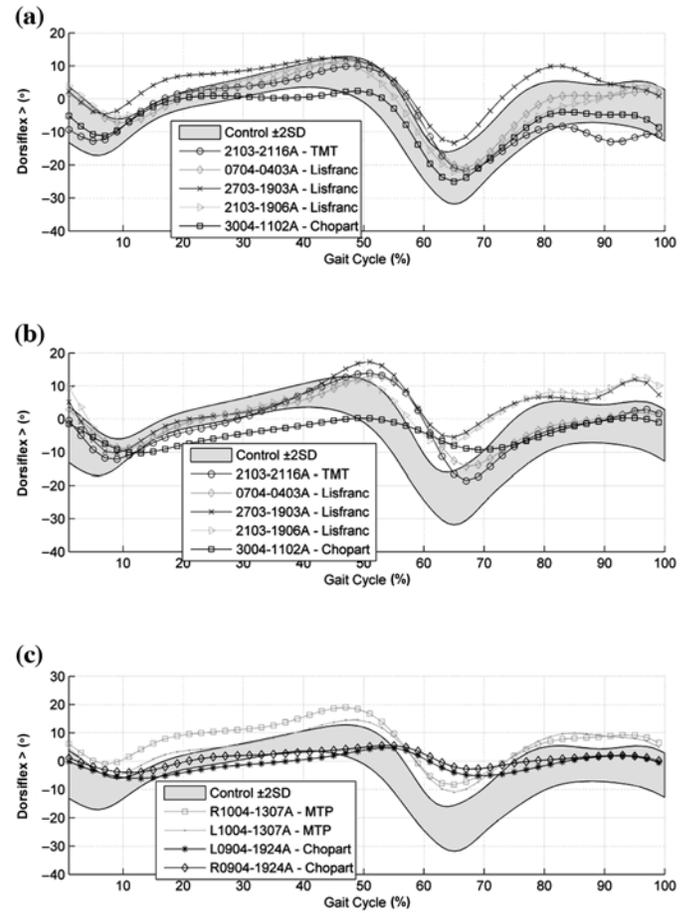
Hip kinematic data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Flex = flexion (positive figures on y-axis), MTP = metatarsophalangeal, TMT = transmetatarsal.



**Figure 5.**

Knee kinematic data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Flex = flexion (positive figures on y-axis), MTP = metatarsophalangeal, TMT = transmetatarsal.

affected limbs of the subjects with TMT and Lisfranc amputation, and peak plantar flexion was reduced compared with the control group as well (Figure 6(b)). The swing phase kinematic patterns seemed consistent with reductions in the peak plantar flexion angle and the need to position the foot appropriately for initial contact (Figure 6(b)). In the subject with MTP amputation, the ankle exhibited a fairly normal movement pattern and range but was simply displaced toward dorsiflexion (Figure 6(c)). The ankle kinematic pattern exhibited by the subjects with Chopart amputation could be characterized by a reduced range of movement and delays in attaining peak angular displacements (Figure 6(b)–(c)).



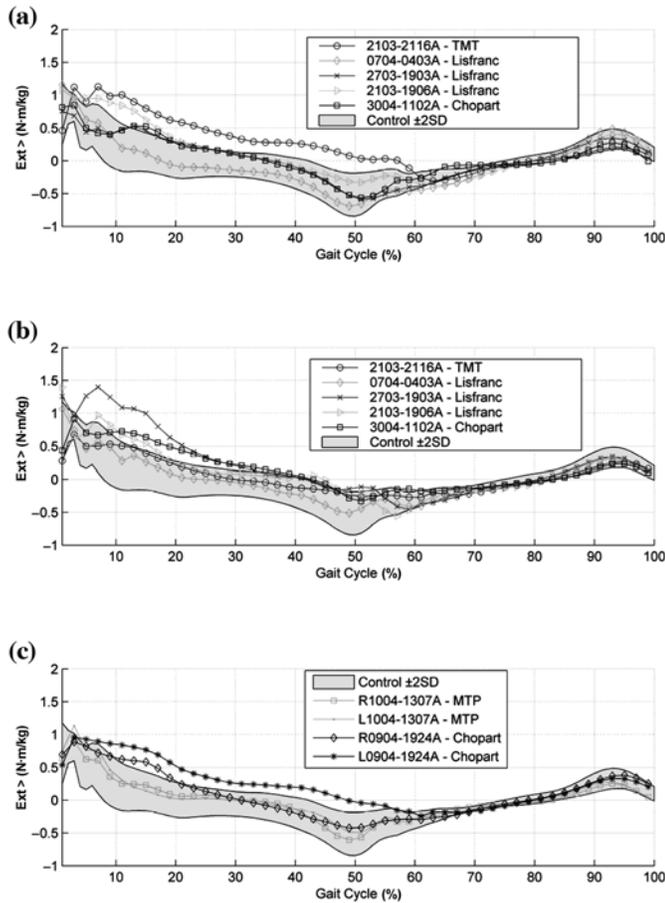
**Figure 6.**

Ankle kinematic data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Positive figures indicate dorsiflexion (Dorsiflex). MTP = metatarsophalangeal, TMT = transmetatarsal.

## Kinetic Data

### Moments

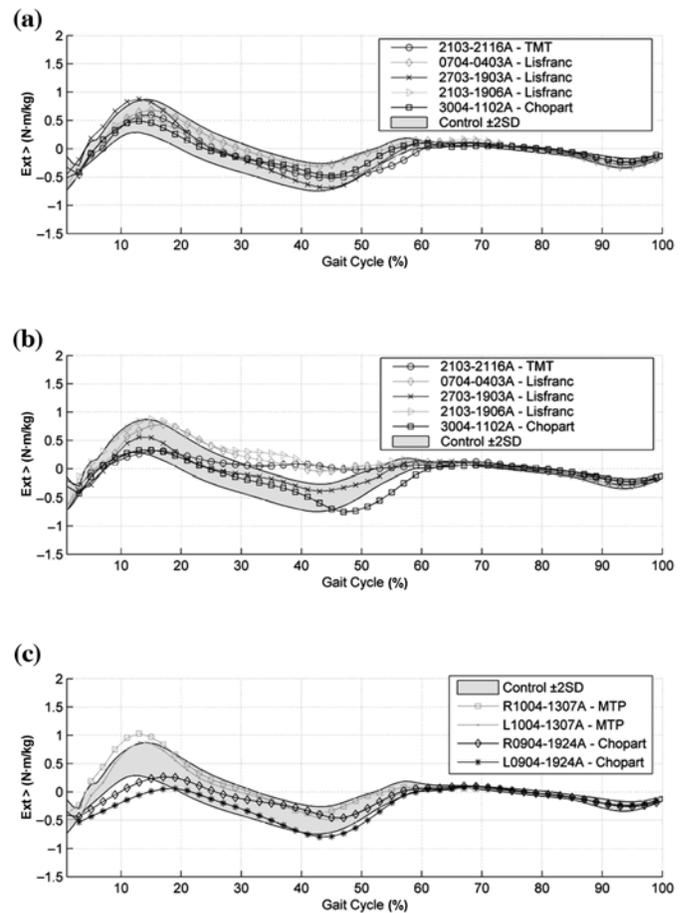
The basic pattern of the hip moments for both the sound and affected limbs was quite variable, with many individuals maintaining an extension moment about the hip joint well beyond midstance (Figure 7(a)–(c)). The hip moment peak associated with loading response was poorly defined in many individuals as well as markedly increased compared with controls (Figure 7(a)–(c)). The timing of the hip flexion moment peak was delayed until after contralateral heel contact on nearly all affected limbs except for the subject with bilateral MTP amputation (Figure 7(b)–(c)).



**Figure 7.**

Hip moment data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data have been normalized by body mass and expressed in newton meters per kilogram. Ext = extension moment (positive figures on y-axis), MTP = metatarsophalangeal, TMT = transmetatarsal.

The knee moment patterns observed on the sound limb were comparable to controls, as were those of the subject with MTP amputation (Figure 8(a) and (c)). On the affected limbs of the subjects with TMT and Lisfranc amputation, a normal knee moment pattern was observed until just after foot flat (Figure 8(b)), after which an extension moment was maintained until about 40 percent GC, when the magnitude of the moment was close to zero. In the subjects with Chopart amputation, the affected limbs exhibited a normal knee moment pattern but the magnitude of the knee flexion moment was increased in two of three cases (Figure 8(b)–(c)).



**Figure 8.**

Knee moment data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data have been normalized by body mass and expressed in newton meters per kilogram. Ext = extension moment (positive figures on y-axis), MTP = metatarsophalangeal, TMT = transmetatarsal.

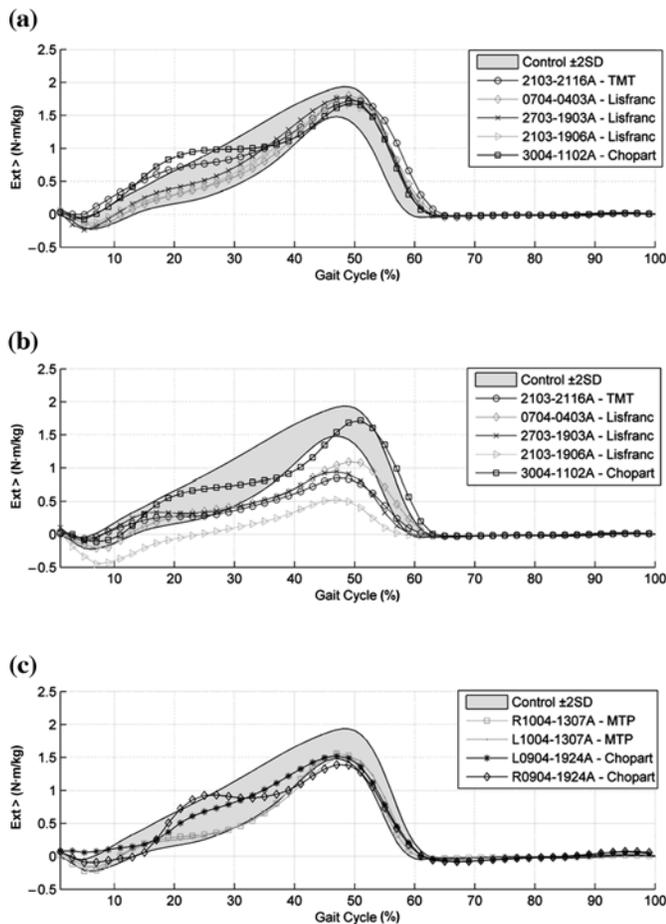
The ankle moments observed on the sound limb were, by and large, similar in pattern and magnitude to those of the control sample (Figure 9(a)). In the subject with MTP amputation, the peak moment was comparable to that observed at the lower 95% CI of the control sample (Figure 9(c)). The subjects with TMT and Lisfranc amputation exhibited a peak plantar flexion moment of between one-third and two-thirds that of the control group (Figure 9(b)). On the affected limbs of the subjects with Chopart amputation, the magnitude of the ankle moment did not increase linearly during the middle of stance phase as in the control subjects but the peak plantar flexion moment was comparable to controls (Figure 9(b)–(c)).

### Powers

On the sound limb of most of the subjects with unilateral amputation, increased hip power generation was observed during early stance, with the magnitude of power generation during terminal stance comparable to the controls but delayed (**Figure 10(a)**). On the affected limbs, power generation during early stance was more variable, with some individuals exhibiting an increase in power and others exhibiting a fairly normal pattern and magnitude of hip power (**Figure 10(b)**). During terminal stance, power generation observed on the affected limbs was comparable

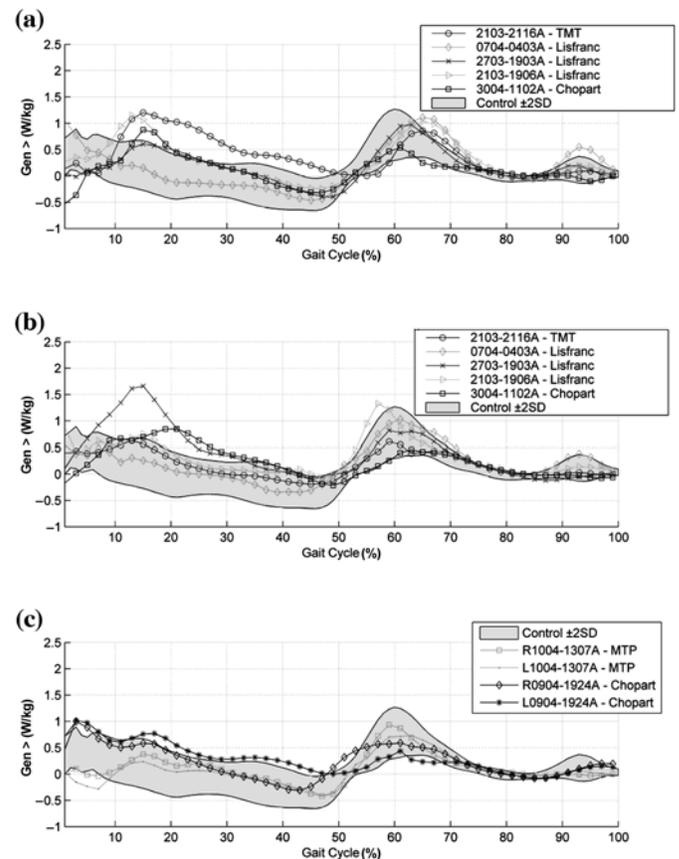
to controls both in terms of timing and magnitude in most subjects with amputation (**Figure 10(b)–(c)**).

Knee power generation and absorption patterns were comparable to controls on the sound limbs of the subjects with unilateral amputation (**Figure 11(a)**). Variable patterns of power generation/absorption were observed on the affected limbs of the amputee cohort, particularly during the power absorption phase following loading response and during terminal stance (**Figure 11(b)–(c)**). Less than normal power absorption was observed in the



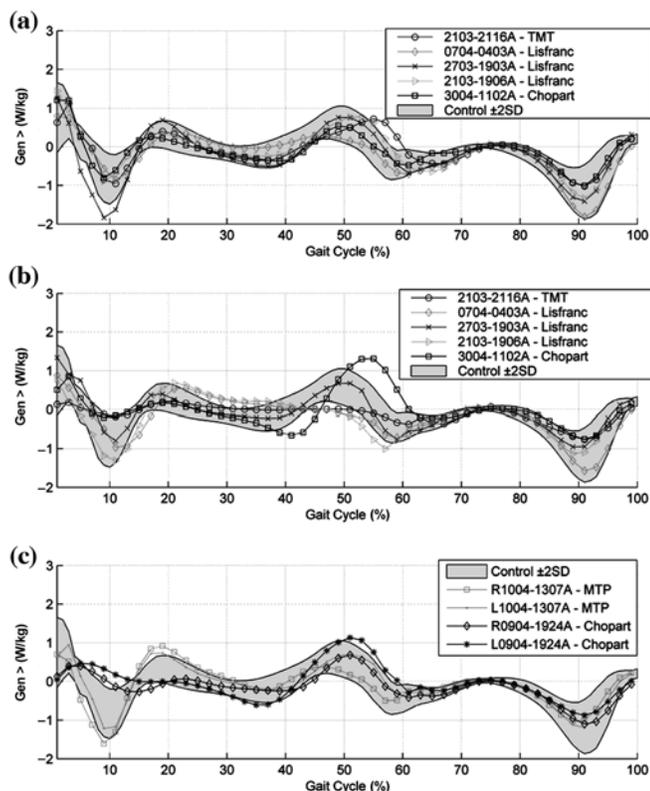
**Figure 9.**

Ankle moment data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data have been normalized by body mass and expressed in newton meters per kilogram. Ext = extension moment (positive figures on y-axis), MTP = metatarsophalangeal, TMT = transmetatarsal.



**Figure 10.**

Hip power data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data have been normalized by body mass and expressed in watts per kilogram. Gen = power generation, MTP = metatarsophalangeal, TMT = transmetatarsal. *Source:* Figures are copyrighted by and reprinted by permission from American Orthopaedic Foot and Ankle Society Inc and were originally published in Dillon MP, Barker TM. Preservation of residual foot length in partial foot amputation: A biomechanical analysis. *Foot Ankle Int.* 2006;27(2):110–116.

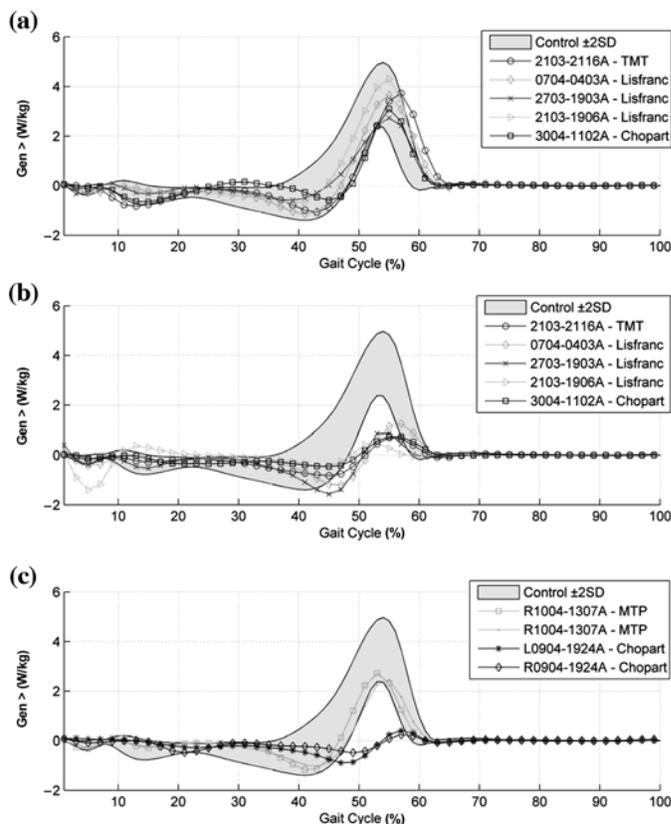


**Figure 11.**

Knee power data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data have been normalized by body mass and expressed in watts per kilogram. Gen = power generation, MTP = metatarsophalangeal, TMT = transmetatarsal.

subjects with Chopart amputation as well as the subject with TMT amputation during early stance but was comparable to controls in the remainder of the amputee cohort (**Figure 11(b)–(c)**).

A normal pattern and magnitude of power generation/absorption across the ankle joint were observed on the sound limbs of the subjects with unilateral PFA (**Figure 12(a)**). In the subject with MTP amputation, the peak ankle power generation was at the lower end of the normal range (**Figure 12(c)**), but once amputation compromised the metatarsal heads, power generation across the ankle was virtually negligible irrespective of amputation level (**Figure 12(b)–(c)**). The power generated by the subjects with TMT and Lisfranc amputation was virtually negligible and comparable to that exhibited on the



**Figure 12.**

Ankle power data for (a) sound and (b) affected limbs of subjects with unilateral amputation and (c) both limbs of subjects with bilateral amputation compared with  $\pm 2$  standard deviation (SD) (95% confidence interval) of nondisabled control cohort. Data have been normalized by body mass and expressed in watts per kilogram. Gen = power generation, MTP = metatarsophalangeal, TMT = transmetatarsal. *Source:* Figures are copyrighted by and reprinted by permission from American Orthopaedic Foot and Ankle Society Inc and were originally published in Dillon MP, Barker TM. Preservation of residual foot length in partial foot amputation: A biomechanical analysis. *Foot Ankle Int.* 2006;27(2): 110–116.

affected limbs of the subjects with Chopart amputation (**Figure 12(b)–(c)**).

## DISCUSSION

Three distinct movement patterns were observed, and on this basis, the “Discussion” will be presented in discrete sections looking at the gait of (1) the single subject with MTP amputation who used insoles, (2) the subjects with TMT and Lisfranc amputation as a group who wore

toe fillers and slipper sockets, and (3) the subjects with Chopart amputation who used clamshell prostheses.

### Metatarsophalangeal Amputee Gait with Insoles

For all intents and purposes, the gait of this individual subject with MTP amputation was comparable to that of the persons without amputation in virtually all respects. Although relatively subtle in nature, progression of the CoP was delayed following midstance (**Figure 3(c)**), consistent with a desire to spare the end of the residuum from the peak GRF, as has been described in detail elsewhere [26]. As a result of this subtle gait adaptation, the peak ankle moments during late stance were at the lower end of that observed in the control sample (**Figure 9(c)**), as was peak power generation across the ankle (**Figure 12(c)**).

### Transmetatarsal and Lisfranc Amputee Gait with Toe Fillers and Slipper Sockets

Once the metatarsals were compromised—through TMT or Lisfranc amputation—significant abnormalities became evident that were characteristic of an inability to progress the CoP beyond the end of the residuum commensurate with the peak GRF and an inability to generate power across the ankle of the affected limb(s).

In the subjects with TMT and Lisfranc amputation, the CoP remained at about 40 percent of shoe length (**Figure 3(b)**), well behind the distal end of the residuum (about 58%–65% of shoe length) throughout most of stance. The timing of the peak GRF on the affected limbs occurred prematurely (45% GC) compared with the control sample (**Figures 1(b)** and **2(b)**) and was commensurate with the CoP being located proximal to the end of the residuum (**Figure 3(b)**). The CoP did not progress beyond the end of the residuum until contralateral heel contact (50% GC), when weight could be transferred to the contralateral limb. In this way, the CoP progressed past the distal residuum when the magnitude of the GRF was rapidly declining (**Figures 2(b)** and **3(b)**). This adaptation would be an effective strategy to spare the distal residuum from the extremes of force typically observed during late stance phase.

The limited distal excursion of the CoP commensurate with the peak GRF led to reductions in the ankle plantar flexion moment (**Figure 9(b)**) and the absence of a knee flexion moment following midstance (**Figure 8(b)**). Persons with TMT and Lisfranc amputation may moderate the external moments at the ankle and knee to compensate for atrophy and weakness of the cojoint ankle plantar

flexors and knee flexors. Alternatively, this strategy could be a means of reducing plantar pressure and shear caused by contraction of the soleus and gastrocnemius musculature. Moreover, this adaptation may minimize the residual foot/socket interface pressures caused by loading the toe lever or may compensate for the prosthetic forefoot being too compliant [26].

The consistent and premature timing of the GRF peaks during terminal stance on the affected limb (**Figure 2(b)**) and the increase in the vertical GRF peak on the sound limb during loading response (**Figure 2(a)**) may reflect a “drop” off the front of the prosthetic forefoot and the person with amputation landing more heavily on the sound limb to check the fall, which has been reported in persons with transtibial amputation [27] when the effective foot length is too short or the prosthetic forefoot too compliant.

That significant reductions in stride length and WV were not observed (**Table 3**) was surprising given the limited distal excursion of the CoP. Reductions in WV in a single subject with TMT amputation were due to reductions in stride length, not cadence (**Table 3**), and appear inconsistent with differences in the design of devices (**Table 1**). Other investigations of persons with PFA due to trauma have also reported WV comparable to nondisabled persons without amputation [8,28]. Interestingly, reductions in WV seem more strongly linked to systemic diseases, such as diabetes and vascular insufficiency, than to PFA per se [10].

Significant reductions in power generation were observed across the affected ankle joint during late stance in persons with Lisfranc and TMT amputation (**Figure 12(b)**). Interestingly, the devices provided to these subjects with amputation were designed to allow ankle motion, but these subjects did not generate any more power across the ankle joint (**Figure 12(b)**) than the subjects with Chopart amputation who wore clamshell devices designed to eliminate ankle motion (**Figure 12(b)–(c)**). To compensate for the loss of sagittal plane ankle power, many individuals demonstrated adaptations at the hip joint(s). On the affected limb, when ankle power was virtually negligible during late stance (**Figure 12(b)–(c)**), increased power generation was observed across the contralateral hip during early stance (**Figure 10(a)**). On the affected limb, increased power generation was also observed during early stance in a number of subjects with amputation (**Figure 10(b)**). Both adaptations provide forward impulse, or push the body from the rear; as such, the hip joints become the primary source of

power generation to compensate for the limited power generated across the affected ankle [29].

Of particular interest was the increased dorsiflexion observed consistently on the affected limbs of the subjects with TMT and Lisfranc amputation (**Figure 6(b)**). This observation is quite unique compared to studies of barefoot walking in this group [6,30–31] and may reflect a measurement error associated with deformation of the prosthetic forefoot. Most of the forefeet in these devices were made from soft foams or flexible carbon plates that would deform to some extent as the person with amputation progressed on the forefoot, thus compromising the rigid-body assumption of the kinematic model and reflecting a relative increase in the dorsiflexion angle obtained. An alternative explanation may be that the heel or device can slip within the shoe as the person with amputation progresses onto the forefoot. Given that the markers defining the foot local coordinate system stay with the shoe, relative motion between the residuum/device and shoe is not captured.

Reductions in plantar flexion were also consistently observed in the group (**Figures 6(b)**) and likely reflect the reduction in power generation across the ankle. Understanding why these persons with amputation do not use the available plantar flexor musculature is difficult given the numerous variables likely to impact this aspect of gait. Reductions in ankle power may serve as a useful means of avoiding localized pressures on the front of the residuum or reducing shear forces should the residuum rotate within the device under the influence of muscle contraction. Perhaps the plantar pressure-reduction strategies typically built into these devices were ineffective. Reductions in ankle power may also reflect a learned gait strategy resulting in the sort of triceps surae atrophy typically observed with these amputation levels. If this were the case, even when subjects were presented with a suitable device, the gait pattern would still be governed by the available plantar flexor muscle strength.

### Chopart Amputee Gait with Clamshell Devices

As a result of the clamshell devices fitted to the subjects with Chopart amputation, in whom ankle motion was eliminated, progression to foot flat during loading response was delayed as the shank and foot segments moved synchronously. The reduced and delayed attainment of the peak ankle angles were comparable to investigations of various prosthetic feet in persons with transtibial amputa-

tion [32–33] and reflect the force-deflection characteristics of the prosthetic feet rather than true ankle motion.

The clamshell devices restored the distal excursion of the CoP (**Figure 3(b)–(c)**) such that the peak GRFs (**Figures 2(b)–(c)**) were borne by the prosthetic forefoot well beyond the distal end of the residuum (40% of shoe length). As such, the peak ankle joint moment (**Figure 9(b)–(c)**) and peak knee flexion moment (**Figure 8(b)–(c)**) were comparable to those observed in the control sample. The clamshell devices were constructed with a rigid socket that encompassed the residuum and leg segments such that the device eliminated ankle motion. The forefeet of these devices were made either from the distal portion of a prosthetic forefoot or had a solid laminated section out to the toe-break (**Table 1**). The devices restored the effective foot length, because they incorporated a stiff forefoot capable of supporting the amputee's body mass during loading and a socket that could comfortably distribute the forces caused by loading the toe lever [26]. Immobilizing the ankle meant that the foot segment and tibial shell were rigidly linked, and as such, the device could moderate the moments caused by loading the toe lever. If the device were to allow ankle motion, then the moments caused by loading the toe lever would need to be controlled by the calf musculature; should this not be possible (either through weakness or discomfort on the residual foot caused by contraction of these muscles), then the persons with amputation would likely not be able to load the prosthetic forefoot in this way.

Despite restoration of the effective forefoot length, ankle power generation was negligible in these subjects with amputation (**Figure 12(b)–(c)**) because motion was eliminated. Even if a clamshell device were constructed to allow ankle motion, the person with amputation would require sufficient calf strength to moderate the external ankle moment and drive the foot into plantar flexion. In the subject with unilateral Chopart amputation, the hip joints became the primary source of power generation to compensate for the limited power generation across the affected ankle (**Figure 10(a)–(b)**). Interestingly, significant power increases at the hip joint(s) were not observed in the subject with bilateral Chopart amputation (**Figure 10(c)**) despite the limited ankle power observed bilaterally (**Figure 12(c)**). Perhaps sufficient power was generated through more subtle adaptations, including prolonged power generation on the left limb throughout stance as well as the early period of power generation during terminal stance on the right limb (**Figure 10(c)**).

Overall, temporospatial aspects of gait were variable. Reductions in velocity, stride length, cadence, and the support phase data were observed in the subject with unilateral Chopart amputation but not in the subject with bilateral Chopart amputation (Tables 2–3), which seems difficult to explain on any basis other than individual variability.

### Prescription and Prosthetic Design

The results from this investigation could be used to suggest that, at the MTP level, the goals of intervention will not likely be centered on improving the mechanics of gait. Other objectives for intervention, such as pressure redistribution, were not considered as part of this investigation. Every attempt should be made to preserve the normal motion of the ankle at this level, given that these persons with amputation can generate power at the ankle joint and, for all intents and purposes, walked as did the control subjects.

While clinicians routinely fit below-ankle devices to maintain motion of the ankle joint at the TMT and Lisfranc levels, the benefits in terms of generating power at the ankle during late stance were not realized. Power generation at the ankle was comparable to that observed in individuals using clamshell devices designed to immobilize ankle motion. As such, clinicians should not prescribe below-ankle devices under the assumption that this will allow substantial power generation at the ankle. Perhaps other considerations, such as minimizing the likelihood of ulceration, may make above-ankle devices a reasonable consideration even at this level, particularly given that many aspects of gait were normalized. Ankle range may be of benefit in a host of everyday activities. These activities were not considered as part of this investigation on level walking.

The results of this investigation suggest that if the goal of intervention is to restore the effective foot length, then the device should incorporate—

- An extensive tibial shell/socket capable of comfortably distributing to the residuum and leg the interface pressures caused by loading the toe lever. The below-ankle sockets fitted to the subjects with Lisfranc amputation seemed unable to achieve this, but the clamshell devices and, more particularly, the anterior shell of these devices seemed appropriate.
- A forefoot capable of supporting the body mass of the person with amputation. The foam fillers or carbon-fiber foot plates seemed unable to fulfill this requirement. When one is choosing a forefoot, the sort of

stiffness typical of prosthetic feet may prove to be a suitable starting point.

- A relatively stiff connection between the foot and leg segments to help moderate the moments caused by loading the toe lever. Either a locked ankle, dorsiflexion stop, or the sort of stiffness inherent in a prefabricated carbon-fiber ankle-foot orthosis may be appropriate [28].

The clamshell devices fitted to the subjects with Chopart amputation seemed capable of restoring the effective foot length and compensating for the limited power generation across the ankle joint—thus normalizing many of the anomalies of PFA gait. While simply installing an ankle joint may seem tempting, doing so will likely eliminate the ability of the device to control the external ankle moments (or compensate for the limited ankle work) and thus require the calf musculature to provide this control. In the absence of sufficient plantar flexor muscle strength or where contraction of the plantar flexors causes other problems like excessive pressure/shear on the distal end, persons with amputation will still not be able to control the external ankle moment and will not adopt a gait pattern in which they load the forefoot and generate power across the ankle. For all intents and purposes, persons with Chopart amputation who use a clamshell device with a free joint will probably walk like persons with Lisfranc or TMT amputation. Articulating the joint with a dorsiflexion stop will likely produce a device that can control the external moments caused by loading the toe lever, but the persons with amputation will still not have the plantar flexor strength to generate power across the joint. As such, the adaptations observed at the hip and knee will likely remain. The effect of training the plantar flexors, in the presence of a suitable device, has yet to be explored.

For any particular client, clinicians must balance multiple goals in creating their prescription, including cosmesis and pressure redistribution, which were not considered in this study. However, the insights gained in terms of prescription should provide valuable information when one considers how the objectives of gait are weighed for any given client.

### Future Investigations

Like most investigations of PFA gait, this study was able to recruit just a small number of participants despite casting a relatively large net. The number of participants reflects 12.5 percent of all PFAs registered with the QALS, and as such, the sample reflects the larger population to a degree. The small number of participants may

reflect the small number of persons with PFA who seek or are referred for prosthetic and orthotic intervention, particularly in light of the total number of amputations performed annually. Investigators may need to explore other avenues for recruitment, including diabetic foot clinics or direct contact with vascular surgical units. One should note that subjects recruited as part of this investigation did not have diabetes or vascular disease. As such, the representativeness of these data to those with PFA due to diabetes and/or vascular insufficiency should be kept in mind.

Investigators may wish to consider a priori grouping of subjects with similar gait patterns irrespective of differences in amputation level or prosthetic intervention. Results from this investigation suggest that the gait patterns of persons with both TMT and Lisfranc amputation (using below-ankle devices such as insoles, toe fillers, or slipper sockets) could, at least in terms of the kinematics and kinetics of gait, be considered fairly comparable. Similarly, persons with amputation using a clamshell device will probably all ambulate much the same irrespective of amputation level because the device dominates the mechanics of gait and differences in amputation level will become irrelevant when encased in the clamshell device.

Multicenter research seems necessary to collect sufficient data to provide meaningful subject numbers. Such research will improve the power of the results and allow more subtle adaptations to be identified.

A range of different investigations are necessary to continue to progress our understanding of PFA gait. Investigators may wish to consider exploring the potential measurement errors affecting the ankle kinematic pattern or teasing apart the many confounding variables that may restrict the ability of persons with PFA to generate power across the ankle joint. While gait research is a necessary step toward a better understanding of the effects of amputation and the influence of prosthetic fitting, basic gait analyses such as this one would benefit from synchronous measurements of trunk excursion and plantar pressure measurement, which would help paint a more complete picture of the underlying causes of the movement patterns exhibited.

## CONCLUSIONS

In conclusion, the purpose of this investigation was to more completely describe the gait of a group of persons

with PFA, with a view to better understanding the underlying mechanical adaptations to amputation and prosthetic fitting.

Major mechanical adaptations were observed once the metatarsal heads were compromised. Persons with TMT and Lisfranc amputation who used devices such as slipper sockets and insoles adopted a gait pattern that limited the distal excursion of the CoP. This adaptation may be a useful means of sparing the end of the residuum from extreme force, moderating the requirement of the triceps surae, or compensating for the compliance of the prosthetic forefoot. The hip joints became one of the major sources of work to compensate for the limited power generation across the affected ankle in persons with TMT and Lisfranc amputation. Persons with Chopart amputation used clamshell-type devices that incorporated a stiff forefoot, rigid ankle, and large anterior leg shell. These features of the prosthetic design allowed relatively normal excursion of the CoP that normalized the knee and ankle moments. Power generation across the affected ankle was still negligible, given that the clamshell socket eliminated ankle range. As such, the hip joints were still a primary source of power generation.

Many of these insights are challenging to contemporary clinical practice. While clinicians often strive to preserve ankle range when designing prostheses, little benefit may exist in terms of the mechanics of gait for persons with TMT and Lisfranc amputation. Given the likelihood of complications such as ulceration, perhaps above-ankle designs may be more readily considered because the benefits associated with maintaining ankle range were not widely realized.

These sorts of insights hopefully will provide clinicians with valuable information when considering how the objectives of gait are weighed against those of cosmesis or pressure redistribution for any given client.

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