**Prosthetics and Orthotics: A Review of Literature**

**Purpose**

This review concerns the use of gait analysis in assessing function of lower limb prosthetics and orthoses. The purpose of this review is to determine:

1. What key variables are measured when performing motion capture experiments to assess the performance of prosthetic or orthoses-assisted gait

2. Are these measurements amenable to 2D video analysis, and if so, what camera view is most useful and what level of accuracy is required to infer differences due to fitting of the prosthesis

3. If a modification to the prosthesis is made, how long before gait changes become evident and are these changes long lasting?

**Introduction**

Regaining walking ability is one of the most prominent goals in the rehabilitation program following lower-limb amputation. Walking ability in this context relates to a natural, symmetric, sustainable gait. The ability to monitor walking ability and evaluate gait patterns during the fitting and rehabilitation process is fundamental to achieving these goals. Walking function can be assessed qualitatively through visual assessment, recall questionnaires, or video. Quantitative measures of gait typically occur in a laboratory setting whereby a motion capture system (optical or accelerometry-based) captures the movements (kinematics) of the individual. Other tasks of daily living can be assessed, including stair climbing, squatting, and rising out of a chair. It is worth noting here that there is an increasing need to assess gait and function outside of the laboratory, in realistic everyday environments, monitoring both quantity and quality of gait.

Kinematics can be coupled with measurement of external forces (traditionally ground reaction forces from a force plate) to determine joint moments and internal forces (kinetics). Kinetic measures are useful for inferring muscle and joint function and these measures combined with electromyography (EMG), can be used to assess which muscle groups are responsible for producing a specific gait pattern. Instrumented prostheses also exist, which enable direct measurement of the forces placed on the amputated limb. Recently, more attention has focused on quantifying pressure and stresses at the interface of the prosthetic socket using pressure sensitive film or complex modelling methods such as finite element analysis.

In reviewing this literature, it is important to keep in mind **three C’s of prosthetic fitting – Comfort, Control, and Cosmetics**. The end goal of the prosthetist is to provide their client with a functional device that will enable him/her to continue their activities of daily living without pain or discomfort and with minimal risk of falling. Despite the improvement in technologies with which to quantify gait and evaluate prosthetic fitting, quantitative data fails to predict the clients’ subjective assessment. Hafner and colleagues (2002) in their review of literature state:

> Unfortunately, most of the biomechanical research performed in the past two decades provides little evidence to support the subjective opinion of enhanced performance.

A more recent quote from Trower (2006) also highlights how modern technologies and our understanding of prosthetics have failed to make an impact on the fitting process:

> The fundamental principles of total contact, uniform pressure distribution, elimination of shear stress and focal pressures, and restoration of correct limb length have been known for a long time. The past decades have seen many studies of socket pressures, vascular flow, muscle activity, tissue density, measurement techniques, fabrication materials, and methods. None of these studies has shown a way to consistently produce a comfortable, stable fit and alignment without the need for trial-and-error fittings.
It is clear that subjective feedback from the client must play a role in any gait analysis and prosthetic fitting and quantitative data alone will not necessarily lead to a good outcome. Rietman and colleagues (Rietman, Postema, & Geertzen, 2002) support this opinion and in their review of literature state:

“It is the authors’ opinion that the future of instrumented gait analysis in prosthetics is mainly for scientific research in which one should critically consider the number of patients to be studied, to show some clinically relevant differences.”

In the most recent review of quantitative gait analysis for prosthetic fitting, Gard (2006) states the following:

“For the time being, quantitative gait analysis may be best used in the research laboratory as opposed to the clinic, but it is important that we continue to strive to effectively integrate these measurements with the experience and skill of the prosthetist and the subjective feedback of the prosthetic user.” (Gard, 2006)

One reason why quantitative metrics do not necessarily match with subjective feedback is that quantitative analysis typically occurs in constrained laboratory settings and does not reflect the environment in which the client will typically be exposed.

“Despite the sizeable history of comparative prosthetic literature and continued analysis of prosthetic components, the link between clinical experience and scientific evidence remains largely unexplored. Acknowledging and targeting areas of perceptive significance will help researchers develop more structured protocols for energy storage and return prosthesis evaluation as well as provide clinicians with information needed to enhance the appropriateness of their clinical recommendations. Expanding test environments to measure activities of perceived improvement such as high-velocity motions, stair ascent/descent, and uneven ground locomotion will provide a more appropriate assessment of the conditions for which energy storage and return prosthetic feet were designed.” (Hafner, Sanders, Czerniecki, & Fergason, 2002)

Finally, the challenge facing today’s modern prosthetist is summarized well by the following statement of Bedotto (2006):

“The current dilemma, given the fact that technology has improved dramatically, is that time is now limited. Lacking is the integration and coordination of treatment necessary to realize the potential of modern technology. As in the past, outcomes will remain limited, unless clinicians devote the necessary time to the provision of total treatment. New advances in treatment must be introduced to coincide with the new advances in prosthetics and orthotics.” (Bedotto, 2006)

**What key variables are measured to assess amputee gait**

**Kinematic Measurements**

Kinematic measurements of amputee gait include temporal-spatial characteristics, joint angles, angular velocities and angular accelerations. Standard temporal-spatial characteristics include:

<table>
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<th>Variable</th>
<th>Unit</th>
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<tr>
<td>Walking speed</td>
<td>Cadence</td>
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<tr>
<td>Stance time</td>
<td>Swing time</td>
</tr>
<tr>
<td>Stride length</td>
<td>Step length</td>
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<td></td>
<td>Stance width</td>
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Self-selected walking speed is generally considered to be the best indicator of a person’s walking ability compared to other quantitative gait measures (Gard, 2006). Walking speed is typically taken using timing gates or a motion capture system for over ground walking or using a treadmill. In able-bodied individuals, the intra-subject variability in self-selected walking speed between days is as much as 7% of the mean. It is not known what variability in walking speed exists in an amputee population although most studies report non-significant differences in self-selected walking speeds when comparing different prosthetic designs (Hafner, et al., 2002). Training can have a large influence on walking speed, with increases reported of up to 48% after a 6 week period (Gauthier-Gagnon, Gravel, St-Amand, Murie, & Goyette, 2000).

Cadence is a measure of the steps per unit time (typically steps/min) and is often reported in conjunction with walking speed. Like self-selected walking speed, cadence has not been shown to significantly change with different prosthetic designs (Hafner, et al., 2002). Day-to-day intra-subject variability in cadence is also high (~3.4%) making subtle comparisons of cadence difficult.

Stride length is the distance between subsequent heel strikes of the same limb and therefore includes the step lengths of both the prosthetic and sound limb. A lengthened stride might indicate an increase in step length in the sound limb and may be a function of improved flexibility of recent prosthetic foot designs. Day-to-day variability in stride length for able-bodied subjects is ~3%. It is interesting to note that one goal of prosthetists and researchers is to achieve a symmetric gait. However, improved function of prosthetic feet may be counter-productive to this goal, allowing greater step lengths and increased walking speed at the cost of decreased symmetry (Hafner, et al., 2002). It is also interesting to note here that individuals, when asked to rank the functions of their prosthetic limb, would much rather have improved walking speed compared to a symmetric gait (Postema, Hermens, de Vries, Koopman, & Eisma, 1997).

Temporal relationships regarding these regions such as heel contact time, heel support time, mid-stance time and push-off times can be influenced by prosthetic foot design and are important for function. For example, energy saving and returning prosthetic feet show increased heel-only and mid-stance support time as compared to conventional prosthetic feet (Perry, Boyd, Rao, & Mulroy, 1997).

Assessments of symmetric or ‘normal-looking’ gait are also performed by comparing joint kinematics (angles) between the amputated and sound limb and comparisons to non-amputee gait, respectively [refer to (Kadaba, Ramakrishnan, & Wootten, 1990) for normal gait parameters]. Kinematic studies on amputees reveal that amputee gait patterns deviate slightly from able-bodied gait (Bateni & Olney, 2002). These differences can be more pronounced, depending on the location of amputation. However, as stated previously, kinematic symmetry of gait does not necessarily correspond to improved function (Chow, Holmes, Lee, & Sin, 2006). Geil (Geil, 2002) performed a study in which 5 different prosthetists performed a fit on one individual and showed that although measurable differences were found in a static orientation, there were no statistically significant differences in kinematic variables during gait. This illustrates the ability of amputees to adapt to subtle alignment differences and ‘internally optimise’ their gait strategy.

Below are the common lower limb kinematic measurements that are compared in various studies:

Hip Kinematics - For below-knee amputee gait, the hip on the affected limb typically undergoes less range of motion (~13º), whereas the hip excursions on the sound limb increase their range of motion (Bateni & Olney, 2002). This change is most likely attributable to the fact that amputees tend to increase their step length on the prosthetic side.
Knee Kinematics - Amputee gait typically undergoes less knee flexion than able-bodied gait during stance phase. Differences in swing phase knee flexion between sound limb and amputee limb have been reported for transfemoral amputees of ~6°. Peak knee flexion during swing phase is also a common metric that is compared across designs and between able-bodied and amputee gait (Sapin, Goujon, de Almeida, Fodé, & Lavaste, 2008).

Ankle Kinematics – Dorsi-flexion range of motion is a common metric used to compare prosthetic foot types, with significant differences from 5 to 10° reported between energy saving prosthetic feet and standard SACH feet (Powers, Torburn, Perry, & Ayyappa, 1994; Snyder, Powers, & Catherine ...., 1995; Torburn, Powers, Guiterrez, & Perry, 1995). The increased range of motion with energy saving and returning feet is believed to be due to the flexible heel, which in turn produces increased sound limb step length (but reduces symmetry of gait).

Foot Clearance - Minimal height of the hallux (big toe) relative to the ground during swing phase provides an estimate of foot clearance and the potential for tripping. Minimal toe height varies with prosthetic knee designs, from 17, to 30, and 37mm (Sapin, et al., 2008).

Kinetic Measurements

Kinetics encompasses the measurement of body forces, which are a summation of gravitational, inertial, and muscular forces. The summation of these forces is what is measured by a force plate and is termed the ground reaction force. Kinematic data are typically coupled with force data to estimate joint moments (or torques) and forces. The distribution of force across the area of body is called pressure, and pressure measurement devices have also been used in an effort to derive clinically-meaningful data for prosthetic fitting. Researchers have also instrumented prosthetic devices with force measuring equipment (typically strain gauges) to obtain more direct measurements of force applied to the prosthetic and amputated limb.

Ground Reaction Forces – The net ground reaction force can be decomposed into three vectors, representing the vertical, anterior-superior (AP), and medial-lateral (MP) components. The vertical ground reaction force during walking consists of two peaks (Figure 1), representing a weight acceptance or loading phase and a propulsive phase.

The weight acceptance peak on the affected side does not show a trend toward increased or decreased vertical force with various prosthetic devices [range +/- 10%; (Hafner, et al., 2002)]. The same peak on the sound limb is often reported to be less with energy saving prosthetic feet [4/6 papers reviewed report reductions (Hafner, et al., 2002), with a maximum reduction of 19% reported by (Powers, et al., 1994)].

The propulsive peak on the affected side commonly shows no statistical difference between different prosthetic foot types (Hafner, et al., 2002). However, this peak is typically diminished in amputee gait compared with normal gait (Wagner, Sienko, Supan, & Barth, 1987). This is due to a loss of the ankle plantar flexor muscles, which are responsible for producing forward
progression and weight support during the propulsive phase. The sound side propulsive peak force shows a trend of increased force using energy saving prosthetic feet compared with regular prosthetic feet, although these changes are small (~7%, (Powers, et al., 1994; Snyder, et al., 1995).

The anterior-posterior component of the ground reaction force resembles a sine wave and consists of a braking phase in the first half of stance and propulsive phase in the second half (Figure 1). Limited results show no change in the braking forces applied to the affected limb with varying prosthetic feet (Hafner, et al., 2002). Similarly, the influence of foot type on the braking force of the sound limb is unknown. Prosthetic foot selection seems to have an influence on the propulsive force of the affected limb. Energy returning prosthetic feet, by their design, return some energy during this propulsive phase and thus produce greater forces compared to standard prosthetic designs. The selection of prosthetic design does not influence the propulsive force production on the sound limb.

Shock - typically measured using an accelerometer attached to the prosthetic device, shock is intended to quantify the high frequency vibrations (>50 Hz) that get transmitted from the prosthesis to the limb during heel strike. Shock data seem to vary with prosthetic foot design (Lehmann, et al., 1993) and are influenced by walking speed and, potentially, by foot orientation.

Joint Moments (or Joint Torques) - Knowledge of joint moments provides useful information regarding joint function, however, due to the redundancy of our joints (i.e. we have more muscles than degrees of freedom at each joint), it is difficult to partition the net joint moment to the independent contributions of each muscle. Electromyography (EMG) is therefore a useful addition to kinetic analysis as it can qualitatively (and in some instances quantitatively) describe which muscles are active and by how much.

Prosthetic foot type does not seem to have much influence on the joint moments at the hip (Colborne, Naumann, & Longmuir, 1992; Perry & Shanfield, 1993). However, alignment and prosthetic design seem to have a large influence on knee joint moments, with large variation in responses in the literature (Colborne, et al., 1992; Schneider, Hart, Zernicke, Setoguchi, & Oppenheim, 1993). The influence of device design and fitting on ankle joint moments is also likely to vary, depending on the prosthetic being used, and whether or not the device has a fixed axis or rotation.

Energy Cost is another common metric used to compare prosthetic designs. Oxygen cost, or VO2, is the most common metric used and is typically normalised by body mass, or distance travelled, or both. It is interesting to note that there are few studies that show significant reductions in energetic cost with energy saving and returning prosthetic foot designs [only 3/9 papers summarised by Hafner et al., 2002], despite subjective claims of improvement from clients who wear these devices (Postema et al., 1997). This might be due to differences in methodology, particularly the exercise modality chosen to compare prosthetics, as well as the clients underlying cause for amputation (i.e. traumatic vs vascular have different effects on VO2). Subjective assessments of energy cost from users also take into consideration other activities of daily living, which might not be included in the test or measurement of VO2.

Instrumented protheses coupled with small data loggers (nowadays provided by flash memory) are now providing the potential to monitor continuous forces across a period of hours, if not days in order to better characterise the net loading stimulus for various activities of daily living (Frossard, et al., 2008). These systems will no doubt become more popular in the future, although it is not yet clear how they might impact the fitting and clinical prescription of prostheses. It is also not clear which forces are most important for the fitting process.

Pressure Measurement Devices have been used to estimate the load distribution at the interface between the prosthesis and skin in an effort to optimise fit for the reduction of blisters, pressure points, and ulcers. Photoelasticity has also been used to qualitatively assess load distributions and has the potential to be used for prosthetic socket fit in the clinic (Sewell, Vinney, Noroozi,
Amali, & Andrews, 2005). One concern with these techniques is that they do not take into consideration the stresses through the underlying tissues and misdiagnosis can occur (Goh, Lee, Toh, & Ooi, 2005; Portnoy, et al., 2007). Portnoy and colleagues (2007) argue that more complex modelling of the surrounding soft tissues is required and demonstrate the feasibility of a finite element model to estimate tissue stresses for real-time fitting. Although technologically feasible, these systems are still far from being clinically tested and usable.

**Subjective Assessments**

In the literature, scant attention is given to subjective ratings of various prosthetics, personal choice of device and deciding factors concerning that choice. Subjective assessments range from descriptive dialog between the prosthetist and client, to functional assessment questionnaires, and numerical rating scales.

Although *descriptive dialog* can provide a lot of information to the prosthetist regarding general prosthesis performance [e.g. (Torburn, et al., 1995)], this qualitative information is difficult to use in a quantitative, comparative manner and has the potential for bias and a placebo effect.

*Functional assessment questionnaires* are designed to examine and compare prostheses in daily modes of living, concentrating on specific parameters of preference or performance (Menard & Murray, 1989). These questionnaires provide additional detail and complexity, but the data provided still cannot be used to determine clinical or statistical significance.

*Numerical rating scales* provide the most detailed analysis of prosthetic performance and patients perceptions, although only a handful of studies in the literature use this method (Macfarlane, Nielsen, Shurr, & Meier, 1991; Postema, et al., 1997). Postema et al. (1997) combined several questionnaires and asked 10 transtibial ambutees to wear different prostheses and then rank in order of importance the factors concerning a prosthesis. Absence of stump pain and stability while walking (Table 1) were the two most important ranked aspects. Note that cosmetically good walking pattern (i.e. symmetric or ‘normal-looking’ gait) ranked second to last in this list.

<table>
<thead>
<tr>
<th>Aspects</th>
<th>Ranking (S.E. of the mean) (maximum ranking 11)</th>
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<tbody>
<tr>
<td>absence of stump pain (n=9)</td>
<td>8.9 (0.72)</td>
</tr>
<tr>
<td>stability while walking</td>
<td>8.7 (0.56)</td>
</tr>
<tr>
<td>no fatigue during walking</td>
<td>7.4 (0.67)</td>
</tr>
<tr>
<td>possibility to walk fast</td>
<td>7.3 (0.93)</td>
</tr>
<tr>
<td>stability in stance</td>
<td>6.7 (0.86)</td>
</tr>
<tr>
<td>feeling of firm contact with the ground</td>
<td>5.5 (0.86)</td>
</tr>
<tr>
<td>rolling-off in a supple way</td>
<td>5.2 (0.51)</td>
</tr>
<tr>
<td>powerful push-off</td>
<td>4.1 (0.80)</td>
</tr>
<tr>
<td>easy turning on prosthetic leg</td>
<td>3.9 (0.81)</td>
</tr>
<tr>
<td>possibility to climb the staircase</td>
<td>3.5 (0.69)</td>
</tr>
<tr>
<td>cosmetically good walking pattern (n=9)</td>
<td>3.3 (0.96)</td>
</tr>
<tr>
<td>possibility of squatting</td>
<td>1.5 (0.62)</td>
</tr>
</tbody>
</table>

Table 1. Ranking of importance (from 1-10) of aspects in perception of the subjects with standard error of the mean between brackets.

**What key variables are amenable to 2D video analysis?**

Many of the kinematic variables mentioned above are amenable to measurement from digital video. A sagittal plane view of the lower limb will provide peak knee and ankle joint excursions and ranges of motion and temporal-spatial relationships. Hip joint kinematics are typically taken as the thigh relative to the pelvis, using external markers on the anterior and posterior superior iliac spines on the pelvis. Pelvis measurement is more difficult and prone to error using 2D video due to the inability to accurately locate the pelvis anatomical landmarks. Trunk lean relative to vertical can also be a useful measure to indicate postural adaptations to change load. Due to the large mass of the trunk, any acceleration of this segment has a large influence on the direction of the ground reaction force vector. Forward trunk lean, for example, will cause the ground reaction force to point further forward, reducing the applied flexion moment to the knee (and reducing the need for quadriceps activation). This type of gait adaptation is common in pathological gait, when individuals alter their body orientations to minimise joint or muscle stress...
in response to pain. Amputee gait involves many such adaptations to re-organise and optimise individual gait patterns.

Frontal plane views are also useful for picking up lateral trunk motion (Sapin, et al., 2008) as well as lower limb orientation to the ground at contact and stance phase. This view will also highlight hip abduction motion throughout swing phase.

Unless ground reaction force vectors are overlayed on video, it is not possible to intuitively or qualitatively determine joint kinetics from video. One system developed by a group of researchers at the Free University of Amsterdam (SYBAR) has this capability and has been used to study amputee gait. Van Velzen and colleagues (Van Velzen, Houdijk, Polomski, & Van Bennekom, 2005) collected sagittal and frontal plane video with SYBAR to investigate whether effects of prosthetic alignment could be observed. They systematically adjusted the fitting of a trans-tibial prosthesis with 15° increments in all degrees of rotation and then took sagittal and frontal-plane video measurements before and after each adjustment. Despite these large perturbations in prosthetic orientation, walking speed and temporal-spatial relationships were not effected. The only noticeable changes in gait were detected in the medial-lateral ground reaction force. The authors concluded that the low resolution of the SYBAR system as well as subjects’ ability to adapt rapidly to the different prosthetic alignment questioned the usefulness of this system for prosthetic fitting.

Given the findings of Van Velzen et al. (2005) and the summaries from many research articles that suggest instrumented gait analysis only has a role in the research environment, the question must be asked whether or not a 2D video analysis system has a place in the clinic for prosthetic fitting? The answer to this question, in my opinion, is yes. But for reasons other than accurately quantifying gait characteristics. As Bedotto (2002) has mentioned, the modern-day prosthetist is faced with a dilemma. That is, they are limited by time and there is a lack of integration and coordination of treatment necessary to realise the potential of modern technology. Digital video can be considered the central aspect of this integration and coordination. A simple video-based system could document the fitting process, reduce a certain amount of trial and error, facilitate treatment planning, provide video-feedback for training, and act as a communicative tool between the prosthetist, the rehabilitation team, and the client.

**What length of time to gait adaptations occur?**

Gait adaptation to a prosthetic device can be considered an optimisation process, in which the individual internally minimises some criteria by adjusting their movement patterns based on some sensory feedback. The sensory feedback often relates to comfort and function, meaning adaptations often occur to minimise pain or discomfort, and to reduce metabolic cost. To date, little is known about the process of gait adaptation, although clinical experience suggests that amputees have a remarkable ability to adapt to a device over a very short period of time (Van Velzen, et al., 2005). In this regard it is important to consider the amputee physiological system not as a fixed component, but as a flexible, biological one (Hafner, et al., 2002).

Given appropriate feedback, gait adaptations can occur within minutes. A recent experiment performed by the author (as yet unpublished) has used haptic devices (vibration and skin stretch) to provide real-time feedback to an individual regarding their knee joint loads as they walk on a treadmill. Within 5 minutes individuals can adjust their gait patterns such that they reduce their knee adduction moment as much as 50% (Wheeler, Shull & Besier, 2009, submitted to Journal of Biomechanics). Ferris and colleagues (Ferris, Liang, & Farley, 1999) showed that the central nervous system is capable of adjusting lower leg stiffness to maintain the trajectory of the centre of mass within one step when running over compliant surfaces. So certainly the body is capable of adjusting to a prosthetic device in a very short space of time. The greater question is how long do these adaptations last? Bedotto (2006) suggests follow-up
assessment is crucial with six-month check ups initially followed by yearly visits when considered necessary, although these suggestions are based on clinical experience, not scientific outcomes. Certainly, more research is required in this area to understand what changes in gait are long-lasting and how often a follow-up analysis is deemed necessary.

**Concluding remarks**

Despite the improvements in prosthetic technology and our ability to accurately characterise gait, the use of quantitative gait analysis for fitting prosthetics in a clinical setting appears limited. However, given the current state of research and the challenges facing the modern day prosthetist, I believe there is room for a video-based fitting and rehabilitation system in a clinical setting. Several fundamental design criteria should be met in order for such a system to be useful and adapted by others:

1. the system must be easy to use and have the potential to reduce the overall fitting time by eliminating some of the trial and error
2. the complete system must be cost effective for a small clinic setting
3. the system must have qualitative as well as quantitative components and should be able to record a numerical assessment of gait function (e.g. drop down menus with each video to record subjective assessments).
4. there is a need to analyse gait long-term and under different environments. Non-level ground for example, alters the influence of prosthetic alignment (Lin, Wu, & Edwards, 2000).
5. there is a need to provide both a fitting assessment tool AND a training-rehabilitation tool. A tool that would help an individual in the training process would be just as beneficial to the client as the initial assessment tool (e.g. timeWARP).
6. there is a need to provide a tool that bridges the communication gap between the physical therapist and the prosthetist. “The physical therapist and the orthotist/prosthetist may not communicate in a timely manner, or they may not communicate at all. This situation results in different treatment approaches without a unified treatment plan. The continuity and focus of treatment are lacking.” (Bedotto, 2006). What is required is a uniform foundation on which to build a treatment plan. The central part of this should be video, by which the PT and prosthetist can obtain objective information to agree and act upon a unified treatment plan.
7. the tool should also be a way for the prosthetist to communicate with the client, supporting the clinical decision making and leading the way to evidence-based practice.
Bibliography


