Medical Engineering and Physics 000 (2017) 1–7

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Technical note

Mechanical characterization and comparison of energy storage and return prostheses

Stacey M. Rigney [a], Anne Simmons [b], Lauren Kark [a,*]

[a] Graduate School of Biomedical Engineering, University of New South Wales, Sydney, NSW 2052, Australia
[b] School of Mechanical and Manufacturing Engineering, University of New South Wales, Sydney, NSW 2052, Australia

ARTICLE INFO

Article history:
Received 7 April 2016
Revised 13 October 2016
Accepted 1 January 2017
Available online xxx

Keywords:
Finite element analysis
Lower-limb amputee
Mechanical testing
Prosthetic foot analysis
Gait analysis

ABSTRACT

The suitability of finite element analysis (FEA) for standardizing the mechanical characterization of energy storage and return (ESAR) prostheses was investigated. A methodology consisting of both experimental and numerical analysis was proposed and trialed for the Vari-flex® Modular™, Flex-foot Cheetah and Cheetah Xtreme by Össur® and a 1E90 Sprinter by Ottobock®. Gait analysis was conducted to determine suitable orientation angles for non-destructive testing (NDT) of the ESAR prostheses followed by a quasi-static inverse FEA procedure within COMSOL Multiphysics®, where the NDT conditions were replicated to determine the homogenized material properties of the prostheses. The prostheses' loading response under bodyweight for an 80 kg person was then simulated, using both Eigenfrequency and time-dependent analysis. The apparent stiffness under bodyweight was determined to be 94.7, 48.6, 57.4 and 65.0 Nmm^{-1} for the Vari-flex® Modular™, Flex-foot Cheetah, Cheetah Xtreme and 1E90 Sprinter, respectively. Both the energy stored and returned by the prostheses varied negatively with stiffness, yet the overall efficiency of the prostheses were similar, at 52.7, 52.0, 51.7 and 52.4% for the abovementioned prostheses. The proposed methodology allows the standardized assessment and comparison of ESAR prostheses without the confounding influences of subject-specific gait characteristics.

© 2017 IPEM. Published by Elsevier Ltd. All rights reserved.

1. Introduction

Energy storage and return (ESAR) prosthetic feet are designed to emulate the compliant structures of the anatomical lower-limb via a spring-like construction of carbon fiber [1]. There has been recent debate over whether ESAR prostheses give lower-limb amputee athletes an advantage [2–4], despite lower-limb amputation generally being associated with unfavorable factors such as asymmetric gait [5] and compensatory mechanisms due to muscle loss [6]. Therefore, any overall advantage would be reliant on these disadvantages being outweighed by the benefits associated with the highly elastic mechanical properties of the ESAR prosthesis and the invulnerability of such properties to muscle fatigue, overtraining or injury.

A universally accepted method of characterizing the mechanical properties of ESAR prostheses has not yet been established, preventing the robust assessment and comparison of such prostheses. Existing mathematical models, including link-segment models [7–9], lumped-parameter models [10,11] and finite element models [12–15], exhibit various limitations in their current forms. Link-segment models are borrowed from able-bodied gait analysis, where joint kinetics and energetics are calculated using inverse dynamics [16]. When applied to amputee gait, the accuracy of these conventional link-segment models is reduced since the underlying rigid body mechanics assumptions are violated by the deformable nature of ESAR prostheses [17,18]. As such, viscoelastic lumped-parameter and finite element models of ESAR prosthesis behavior have been proposed as alternatives. Both techniques first incorporate *in vitro* mechanical testing of the prosthesis at various orientation angles and attempt to replicate these boundary conditions in the relevant model [10–13]. Consequently, the accuracy of these models is highly dependent on the specific conditions under which they were developed. Mechanical testing protocols have not yet been developed for this purpose, as existing procedures focus on safety or generic stiffness categorization [19–24], and often permit sliding of the distal end (or contact point) of the prosthesis. This is problematic, as the overall stiffness of a prosthesis is dependent on both its orientation angle and the displacement of its distal end [25] so unless both degrees of freedom are included in the subsequent mechanical model then the results obtained from such methods are not truly indicative of the prosthesis' mechanical behavior [26].

* Corresponding author.
E-mail addresses: stacey.rigney@unsw.edu.au (S.M. Rigney), lauren.kark@unsw.edu.au (L. Kark).

http://dx.doi.org/10.1016/j.medengphy.2017.01.003
1350-4533/© 2017 IPEM. Published by Elsevier Ltd. All rights reserved.

The lack of standardized techniques to measure prosthesis behavior results in the inability to benchmark performance without the confounding effects of subject-specific gait mechanisms. As such, the purpose of this paper is to propose a new method for characterizing the mechanical behavior of ESAR prostheses that uses a combination of experimental and numerical techniques. We hypothesize that finite element analysis (FEA) is a viable method for standardizing the assessment of ESAR mechanical performance as it allows the control of variables such as bodyweight, friction and orientation.

2. Methods

The mechanical characteristics of an unused Vari-flex® Modular™, Flex-foot Cheetah and Cheetah Xtreme by Össur® (Reykjavik, Iceland) [27] and a used 1E90 Sprinter by Ottobock® (Duderstadt, Germany) were determined using both experimental and numerical analysis (Fig. 1). The homogenized material properties of the prostheses were first calculated using the inverse FEA procedure proposed in a preliminary version of this study [28], which consisted of amputee gait analysis to determine the appropriate sagittal plane orientation of the prostheses during non-destructive testing (NDT), and the replication of this NDT in COMSOL Multiphysics® 5.0 (Stockholm, Sweden). The material properties were then used in time-dependent FEA simulations of each prosthesis' loading response under gravity for a unilateral lower-limb amputee with a nominal body mass of 80 kg (standing only on their prosthetic leg), where throughout this paper 'body mass' is defined as the total mass of the amputee athlete using the prosthesis, excluding the mass of the prosthesis itself. Damping properties for this simulation were determined using a conjunction of Eigenfrequency analysis and an optimization routine based on the conservation of energy principle, which will later be discussed in detail. The final outputs of the methodology were the calculation of the energy stored and dissipated by each prosthesis as well as their overall stiffness and efficiency.

2.1. Experiments

A female right-sided transtibial amputee athlete using a Cheetah Xtreme volunteered for the study in accordance with UNSW Australia Human Research Ethics Committee approval HC13054. The subject sprinted at maximum speed along an indoor 100 m running track instrumented with 20 MX T40-S 250Hz Vicon® (Denver, Colorado, USA) motion capture cameras. As described in the preliminary version of this study, reflective markers were placed along the lateral side of the prosthesis in order to determine the nominal range of sagittal plane angles θ that the longitudinal axis of the prosthesis made with the laboratory vertical during gait (for further detail see Fig. 2 of [28]). The mean resulting angles were determined to range from 12° to 22° during initial and mid-stance, increasing up to a maximum of 52° during terminal stance.

For longitudinal axis orientations of $\theta = 12, 22°$, each prosthesis underwent constant strain rate loading of $\dot{y} = \pm 1 \text{mms}^{-1}$ up to a maximum load of 1000N using a uniaxial displacement-controlled Instron® (Norwood, Massachusetts, USA) material testing system, where a positive crosshead velocity ($\dot{y} > 0$) indicates the prosthesis was being loaded in compression. Stress-relaxation testing of each prosthesis was also conducted by holding a constant displacement of $y = 40$ mm for 350 s, at the 12° orientation only. The crosshead was fitted with a 5000 N load cell and the induced vertical reaction force was measured at a sample rate of 1000 Hz.

During loading and unloading, static friction between the distal end of the prosthesis and the load cell prevented relative motion until the magnitude of the horizontal reaction force of the prosthesis was greater than the opposing frictional force. Consequently, the horizontal force F_x was calculated from the measured vertical force F_y during periods of 'slip' by assuming maximum possible friction Eq. (1), where the coefficient of static and dynamic friction were assumed to be equal and therefore both denoted by μ. During periods of 'stick' the horizontal force F_x was assumed to vary linearly with load cell displacement (Fig. 2).

$$F_x = -\mu F_y \operatorname{sgn}(\dot{y})$$
$$\text{where } \operatorname{sgn}(\dot{y}) = \begin{cases} +1 & \text{if } \dot{y} > 0 \\ -1 & \text{if } \dot{y} \le 0 \end{cases} \tag{1}$$

2.2. Numerical model

2.2.1. Quasi-static inverse FEA

The material properties of the prostheses were determined by reproducing the constant strain rate testing conditions within a quasi-static FEA model and finding the parameters for which the experimental and numerical results were equivalent. The 3D geometry of each prosthesis was created in SOLIDWORKS® 2014 (Waltham, Massachusetts, USA) using 3D laser scans that were validated using Mitutoyo® (Kawasaki-shi, Kanagawa, Japan) digital Vernier calipers. Using a 3D Solid Mechanics interface and Multifrontal Massively Parallel Sparse (MUMPS) direct parametric stationary solver within COMSOL Multiphysics®, the load cell was modeled as a rigid body and each prosthesis as a linearly viscoelastic, isotropic and homogenous body, where Poisson's ratio was assumed to be 0.3 based on previous ESAR prosthesis studies [12–14]. A homogenized model of the carbon fiber was adopted since the results of importance belonged to the global prosthesis behavior as opposed to the small-scale mechanical characteristics within the material, including prosthesis reaction force and energy storage. For each prosthesis case, a contact pair was defined between the surface of the load cell (source) and the distal end of the prosthesis (destination) using the coulomb friction model described in Eq. (1). The boundary where the support block attached to the prosthesis was fixed, a parametric prescribed displacement of $y = \pm 1$ mm was given to the load cell domain for each quasi-static time-step, and all other boundaries remained free.

The effect of spatial discretization on the calculated reaction force was assessed using the above boundary conditions and arbitrary parameters of $E = 60$ GPa, $\mu = 0.1$ and $y_{\max} = 10$ mm. During this convergence study, all simulations utilized a tetrahedral mesh and the maximum element size was adjusted from 0.0416 to 0.00284 mm at increments of approximately 0.001 mm. Consequently, element sizes were chosen to ensure that the absolute value of the difference between the calculated F_y for the selected mesh and the converged mesh was less than 0.05 N (designated as ΔF_y in Table 1).

For each prosthesis, the value of E and μ was determined using a finite element model in conjunction with the reaction force data from their respective 12° experiment. The value of E was first estimated using $\mu = 0$ by implementing a frictionless contact pair between the load cell and the prosthesis and minimizing the square of the difference between the target and simulated vertical reaction force. Since this simulation was frictionless and yet the experiment was not, the target reaction force was defined as the measured vertical reaction force F_y at the position during the 'stick' section of the unloading phase when the horizontal reaction force F_x was momentarily zero, as highlighted by the dashed line in Fig. 2. Once E was known, the optimization simulation was repeated with friction to determine the value of μ. The lower limit, initial guess and upper limit for E were 50, 60 and 70GPa respectively, and for μ these values were 0.0, 0.1 and 1.0 respectively. The calculated values of E and μ were validated for each prosthesis by

Download English Version:

https://daneshyari.com/en/article/5032757

Download Persian Version:

https://daneshyari.com/article/5032757

Daneshyari.com