Ultrasonics 54 (2014) 779–788

Contents lists available at ScienceDirect

Ultrasonics

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

Estimation and visualization of longitudinal muscle motion using ultrasonography: A feasibility study

Jizhou Li [a], Yongjin Zhou [a,b,*], Kamen Ivanov [a], Yong-Ping Zheng [b]

[a] Shenzhen Institutes of Advanced Technology, Chinese Academy of Sciences, China
[b] Interdisciplinary Division of Biomedical Engineering, The Hong Kong Polytechnic University, China

ARTICLE INFO

Article history:
Received 27 March 2013
Received in revised form 31 August 2013
Accepted 27 September 2013
Available online 15 October 2013

Keywords:
Ultrasound
Image processing
Motion estimation
Muscle contraction

ABSTRACT

Ultrasonography is a convenient and widely used technique to look into the longitudinal muscle motion as it is radiation-free and real-time. The motion of localized parts of the muscle, disclosed by ultrasonography, spatially reflects contraction activities of the corresponding muscles. However, little attention was paid to the estimation of longitudinal muscle motion, especially towards estimation of dense deformation field at different depths under the skin. Yet fewer studies on the visualization of such muscle motion or further clinical applications were reported in the literature. A primal–dual algorithm was used to estimate the motion of gastrocnemius muscle (GM) in longitudinal direction in this study. To provide insights into the rules of longitudinal muscle motion, we proposed a novel framework including motion estimation, visualization and quantitative analysis to interpret synchronous activities of collaborating muscles with spatial details. The proposed methods were evaluated on ultrasound image sequences, captured at a rate of 25 frames per second from eight healthy subjects. In order to estimate and visualize the GM motion in longitudinal direction, each subject was asked to perform isometric plantar flexion twice. Preliminary results show that the proposed visualization methods provide both spatial and temporal details and they are helpful to study muscle contractions. One of the proposed quantitative measures was also tested on a patient with unilateral limb dysfunction caused by cerebral infarction. The measure revealed distinct patterns between the normal and the dysfunctional lower limb. The proposed framework and its associated quantitative measures could potentially be used to complement electromyography (EMG) and torque signals in functional assessment of skeletal muscles.

1. Introduction

Many biomechanical investigations require quantification of muscle contraction. It has been shown that different regions of muscle can be activated to complete different motor tasks [1] and to generate forces in different directions [2]. Accurate *in vivo* measurements of muscle motion will expand our understanding of normal and abnormal muscle functioning. Methods that provide comprehensive quantitative description of muscle contractions and allow to correlate the actions of several muscles, including agonistic and antagonistic actions, would allow engineers and doctors to better understand, diagnose and even treat musculoskeletal disorders [3–6]. Muscle contractions are usually studied using electromyography (EMG) [7–11], force and displacement transducers [12,3,13]. However, surface electromyography and other non-invasive methods cannot provide information about morphological changes of contracting muscles, while intramuscular electromyography is invasive method which assumes painful injection and discomfort for the patient.

Many imaging modalities have been used to study muscle functioning. Besides CT, MRI and PET [14–17], ultrasonography (US) has been recognized as one of the most attractive techniques to assess human muscles under both static and dynamic conditions because of its low cost, high flexibility and extraordinary patient friendliness [18]. In the early 1980s, it was first discovered that in ultrasound examination muscles which suffer from disease have different appearance compared with healthy ones [19]. Today, US has been proved to allow accurate measurement of changes in muscle thickness [20–25], fiber length [26,27], pennation angle [20,26,28–30] and cross sectional area [31,10,32].

Some algorithms track important features of contracting muscle from one frame to the next automatically [33–36]. Other

Abbreviations: GM, gastrocnemius muscle; EMG, electromyography; US, ultrasonography; CCV, a pseudocolor view using color coding scheme; MP, motion profile; DAMP, Dorsoventrally Averaged Motion Profile; TAMP, Temporally Averaged Motion Profile; MVC, maximal voluntary contraction; CLG, Combined Local–Global; PSNR, peak-signal-to-noise ratio; RMS, Root Mean Square.

* Corresponding author. Address: 1068 Xueyuan Avenue, Shenzhen University Town, Shenzhen, China. Tel.: +86 755 86392280.

E-mail address: y.zhou.cn@ieee.org (Y. Zhou).

0041-624X/$ - see front matter © 2013 Elsevier B.V. All rights reserved.
http://dx.doi.org/10.1016/j.ultras.2013.09.024

algorithms are based on computation of the muscle motion field called "optical flow" and usually involve computation of various temporal and spatial derivatives of original image sequences [38–44,37]. According to Horn's taxonomy [45], the motion field is the 2D projection of the 3D motion of surface in the real world, whereas the optical flow is the apparent motion of the brightness patterns in a sequence of images. Generally speaking, four groups of methods are described in the literature: differential methods, block-matching methods, energy-based methods and phase-based methods. Overview of these methods can be found in [46–48].

Revell et al. [40] and Shinohara et al. [49] used ultrasound for visualization of muscle contractions and deformed muscles. Cronin et al. [37] developed an automatic fascicle tracking method based on the Lucas–Kanade optical flow algorithm with an affine assumption. Typically it is difficult to perform motion estimation in ultrasound images because of the high level of noise they contain. In addition, due to the complicated local motions of muscles, the requirement of optical flow algorithms for constant intensity cannot be met in many regions of the ultrasound image. There are few works on quantitative measurement of muscle motion based on ultrasound image sequences which aim to interpret synchronous contractions of collaborating muscles. In this study, we propose a novel framework for visualization and analysis of muscle motion using ultrasound. We also propose novel quantitative measures for estimation of muscle motion as a part of the framework. The objective of this study is to perform visualization and quantitative analysis of high-resolution ultrasound image sequences captured from contracting muscles at different depths under the skin and at various temporal points from rest to contraction.

The paper is organized as follows. Section 2 provides a brief survey of the motion estimation methods and the primal–dual algorithm, and an introduction to the proposed framework. The experimental setup is given in Section 3 and results are described in Section 4. In Section 5, the proposed methods are discussed. Finally, concluding remarks are given in Section 6.

2. Methods and materials

2.1. The primal–dual optical flow algorithm

Modeling and estimation of dense optical flow fields have been intensively studied in the literature. Starting with Horn and Schunck [50] and Lucas and Kanade [51], researchers have developed a variety of models for effective flow computation. Here we focus on the variational $TV - L^1$ formulation of the optical flow convex optimization, where the optical flow is estimated as a minimizer of the energy (see e.g. [52]):

$$v = \arg\min_{v} \int_{\Omega} |\nabla v| + \lambda \|\rho(v)\|_1 \tag{1}$$

where $\Omega \subseteq \mathbb{R}^2$ is the image domain, $v = (v_1, v_2)^T$ is the motion field, $\rho(v) = I_t + (\nabla I)^T(v - v^0)$, I_t is the time derivative of the image sequence, ∇I is the spatial image gradient, v^0 is some given motion field and the parameter λ is used to define the tradeoff between data fitting and regularization.

In ultrasound image sequences, it is unlikely, due to illumination changes and shadows, that the image intensities stay constant over time. It should be noted that many researchers are pursuing this ultrasonic echo intensity as a meaningful change due to material property changes in tissue [53]. This motivates the following slightly improved motion estimation, which explicitly models the varying illumination by means of an additive function u (see e.g. [54]):

$$\rho(u, v) = I_t + (\nabla I)^T(v - v^0) + \beta u \tag{2}$$

The function $u : \Omega \to \mathbb{R}$ is expected to be smooth and hence we also regularize u by means of the total variation. The parameter β controls the influence of the illumination term. The improved motion estimation model is then obtained as:

$$\min_{u \in \mathbb{X}, v \in \mathbb{Y}} \int_{\Omega} |\nabla u| + |\nabla v| + \lambda \|\rho(u, v)\|_1 \tag{3}$$

where $\mathbb{X}$, $\mathbb{Y}$ are finite dimensional vector spaces. The details of the primal–dual algorithm [55] used in this study can be found in the Appendix A.

2.2. Proposed framework

After computation of the motion field, a novel framework is proposed to visualize and analyze the muscle contraction. As shown in Fig. 1,

1. The motion field can be illustrated using the established color coding scheme described in [48], where both orientation and magnitude of the motion are taken into account to construct a pseudocolor view (CCV). More details are given in the results section.
2. Inspired by the color coding strategy in Doppler ultrasound [56], we use red color for motion to the right and blue color for motion to the left after the motion field has been averaged along the horizontal axis (from proximal to distal) and plotted vs. vertical axis (from superficial to deep). It should be noted that with such plot we intend to visualize the motion profile (MP) along the vertical axis.
3. Three plots are then derived from MP. The MP for each frame can be averaged along the vertical axis. These averaged values together form the Dorsoventrally Averaged Motion Profile (DAMP), which reflects the temporal changes of the overall contraction level.
4. The MP for each frame is temporally averaged, and these average values form the Temporally Averaged Motion Profile (TAMP), which reflects the contraction ability of the muscle along different depths from skin.
5. MP for all frames can be aligned along temporal axis to construct a temporal panoramic view of the muscle motion.

3. Experiments

3.1. Subjects

Eight healthy male subjects (mean ± SD, age = 33.4 ± 5.2 years; body weight = 63.8 ± 10.3 kg; height = 1.65 ± 0.07 m) volunteered to participate in this study. No participant had a history of neuromuscular disorders, and all participants were aware of experimental purposes and procedures. Human subject ethical approval was obtained from the relevant committee of the Hong Kong Polytechnic University, and informed consent was obtained from the subject prior to the experiment.

3.2. Experimental protocol and data acquisition

The testing position of the subject was in accordance with the Users Guide of a Norm dynamometer (Humac/Norm Testing and Rehabilitation System, Computer Sports Medicine, Inc., Massachusetts, USA). Each subject was required to put forth his maximal effort of isometric plantar flexion for a period of 3 s with verbal encouragement provided. The maximal voluntary contraction (MVC) was defined as the highest value of torque recorded during the entire isometric contraction. Each test was repeated twice with a rest of 5 min between the two subsequent trials. Before the actual test, each subject was asked to perform a