ULTRASOUND BASED ESTIMATION OF MUSCLE CONTRACTION

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ABSTRACT

In the absence of a single satisfactory control path from the amputee to the prosthesis, the search for new alternatives is of great interest. This paper introduces the idea of using ultrasound measurements of human tissue as a basis for generation of control inputs to a prosthesis. One of the methods employed, an optical flow estimation algorithm, is based on the work of others, but a simpler, one-dimensional implementation is developed. The second method presented is based on cross correlation and is demonstrated to yield smooth muscle force estimates under ideal conditions. The results are compared to those obtained with an EMG based technique. Weaknesses are discussed and suggestions for further research are given.

INTRODUCTION

Every command path from human to machine is based on physiological states that can be controlled by the human central nervous system. EMG, myoacoustic signals, voice commands, muscle bulge, force applied to analog sensors or mechanical switches, sensors attached to exteriorized tendons and numerous other means have been utilized for forward paths in powered prosthetic applications. In 1976, Hogan stated: "As yet no completely satisfactory or generally applicable means of providing a proportional forward path has been found, although a considerable amount of work has been done on the problem" [1]. 21 years later progress has been made, but the statement is still valid. The methods and results described in this paper are preliminary and in no respect solve the forward path problem, but they nonetheless constitute a supplement to the numerous methods that are demonstrated to come short.

The present study is based upon previous work by Bertrand, Meunier, Douchet et.al. [2], who demonstrated the ability to estimate the amount as well as the *location* of muscle contraction in a sequence of two-dimensional ultrasound cross-sectional images of a forearm. Their method was based on tracking the ultrasound speckle, a complex and usually unwanted interference pattern inherently present in all ultrasound images of inhomogenous objects. Since under certain assumptions the speckle pattern follows the movements of the object being studied, it can be utilized for tracking tissue movements as well as multi-axial tissue strain; the biaxial strains of a contracting muscle's cross section is an example of the latter. Thus, their technique offers a potential proportional forward path in the above mentioned sense.

Bertrand et.al. tracked the speckle of a two-dimensional image by using an efficient off-line optical flow technique. For a method to be practically applicable to prosthesis control, the components and power source required to implement it in real time must be contained within the envelope of the human body. Novel application-specific integrated cirquits enables much of the high-speed ultrasound sampling and basic signal processing involved to be performed with very small space and power requirements [3], while new battery technologies allow for the prosthesis to carry considerably larger amounts of energy than was the case earlier [4]. Futhermore, the complexity and thus the resource requirements of the method of Bertrand et.al. can be greatly reduced by implementing it in one dimension, e.g. using single ultrasound beam measurements, instead of in two dimensions, at the cost of possible quality and robustness degradation. This paper presents initial results related to such a simple, one-dimensional implementation of the method mentioned, as well as a correlation-based technique, used for estimation of dynamic isometric muscle contractions. While a speckle pattern is the result of

echoes from thousands of microscopic point reflectors (e.g. blood cells), the methods were demonstrated by tracking the echo from a tissue macro structure (bone-soft tissue interface) rather than the plain speckle. The results are compared to those obtained with an EMG based technique

THEORETICAL BACKGROUND

The present problem can be described as follows. With a suitable pulse repetition rate, ultrasound pulses are transmitted into a medium. The echo from the medium is recorded for each pulse, resulting in an echo signal sequence. Inhomogenities in the medium (in this case a human body) cause characteristic patterns, peaks, to appear in the echo signal envelopes, and the position of the peaks within the signal yields information of the depth of the inhomogenities that caused them. Our task is to track the movements of these inhomogenities through the sequence of echo signals.

Optical flow

The optical flow (OF) method used in this study is a one-dimensional version of that of Bertrand et.al. [2], and is based on the assumption that a given signal sequence displays a velocity field that is linear with respect to space. In mathematical terms, for a spatially one-dimensional signal $s(x, t)$, this means that the velocity profile $v(x)$ in the spatial direction can be described by the equation

$$v(x) = ax + b \tag{1}$$

where a and b are constants; b is related to the translational movement of the signal in the spatial direction, while a is related to the strain i.e. the deformation (stretch or compression) of the signal. For spatiotemporally small regions of interest (ROI), this linearity assumption holds.

From the chain rule for derivatives follows that

$$\frac{ds}{dt} = \frac{\partial s}{\partial x}v + \frac{\partial s}{\partial t} \tag{2}$$

The left-hand side of (2) represents the amplitude or intensity change of the moving signal. Bertrand et.al. chose to neglect this term on the assumption that the "interframe deformations" (i.e. the deformation of the signal from one echo signal to the next) are small. With this assumption, (1) and (2) yield:

$$\frac{\partial s}{\partial x}ax + \frac{\partial s}{\partial x}b = -\frac{\partial s}{\partial t} \tag{3}$$

The partial derivatives in (3) can be estimated from the signal sequence for all relevant values of x and t, and the velocity profile parameters a and b can be found by a least squares fit. Now the estimated velocity profile can be integrated to yield accumulated translation (position) and strain.

Cross correlation technique

The cross correlation (XC) technique employed is a straightforward procedure. The cross correlation function of two subsequent signal ROIs was estimated. The index shift of the maximum value of this estimate was taken as a first crude approximation of the spatial translation between the ROIs. Then a parabola was fitted to pass through the maximum point of the cross correlation estimate and its two neighbouring points, and the final spatial translation estimate was taken as the position of the vertex of this parabola. The ROI of the next echo signal was chosen according to the estimated translation in order to always keep two subsequent ROIs maximally correlated (to keep the same pattern within the ROI).

EXPERIMENTAL SETUP

In order to measure ultrasound pulse-echo modulated by a contracting muscle while at the same time recording EMG and actual force developed by the same muscle, the biceps was chosen for convenience. For estimation of static isometric contractions, one used a setup similar to that of Philison and Larsson [5], figure 1. A healthy male volunteer was seated in a wooden chair with his right upper arm vertical, his elbow firmly supported by the chair back support, his forearm horizontal and his hand fully supinated but relaxed. This posture is reported to minimize the contributions of synergetic muscles to the elbow flexion torque; furthermore, findings by Hogan and Mann suggest that under the given conditions, the pattern of activity of a single muscle is representative for the whole group [6].

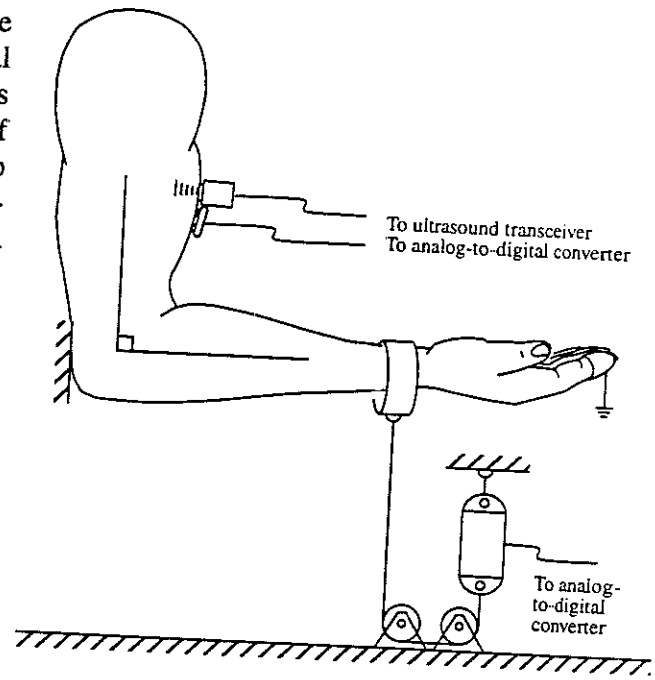

Figure 1: Experimental setup. The ultrasound probe (top) and the EMG electrode were placed over the biceps, while the wrist was strapped to the load cell (lower right).

The test person's wrist was strapped to an analog load cell. The output signal of the load cell was digitized with a sampling frequency of $2kHz$ and passed through a discrete 4. order Butterworth lowpass filter with $20Hz$ cutoff frequency.

For EMG logging one used a Liberty MYO115 research electrode that was placed on the medial side of the biceps, its EMG output signal being fed through a first-order $500Hz(3dB)$ analog low-pass filter and then digitized at $2kHz$. Subsequently it was filtered by a discrete first-order $1Hz(3dB)$ high-pass filter to eliminate any DC offset.

For the ultrasound measurements a Panametrix EPOCH II Ultrasonic Flaw Detector with a $4MHz$ probe was utilized. The probe was mounted on an attachment device that was placed over the biceps of the test person and then strapped firmly to his arm in order to keep it in a constant position during the measurements. The device employed a lockable ball joint to allow for the probe's orientation to be adjusted. In this experiment one chose to point the ultrasound beam towards the humerus, because the soft tissue-bone interface virtually causes total reflection of the incident sound and thus yields a clear and consistent echo. The EPOCH II was set to pulse-echo mode, so that the single transducer could be used both for transmitting the ultrasound into the tissue and for receiving the echo reflected from the tissue structures. The analog radio-frequency (RF) echo output of the EPOCH II was digitized by a digital oscilloscope and downloaded via a GPIB bus to a computer for storage and subsequent analysis. Due to the GPIB data transfer delay, one obtained an effective pulse repetition rate of slightly less that $10Hz$ with a sampling resolution of $8bits$. These parameters are far from optimal and yield a crude signal representation, but proved to be sufficient for demonstration of the present method. For each pulse, the RF echo was sampled at a rate of $100MHz$ for $5\mu s$, yielding 500 data points. This approximately corresponds to a "roundtrip distance" (travelled by the sound) of $7.5mm$. The time lapsed from the emission of the pulse to the start of the $5\mu s$ sampling interval was set on the oscilloscope front panel so that the $3.75mm$ tissue "image" was taken at the desired depth ("range-gating").

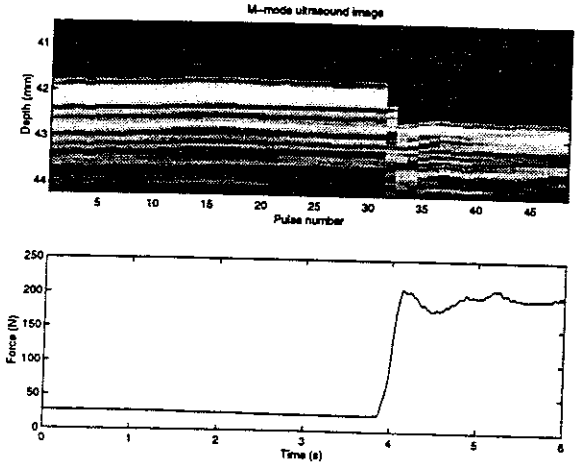

Figure 2: Range-gated M-mode ultrasound image of bone-soft tissue interface while a muscle (biceps) between the ultrasound probe and the bone goes through a sudden contraction (top). Simultaneous force generated by the biceps muscle (bottom).

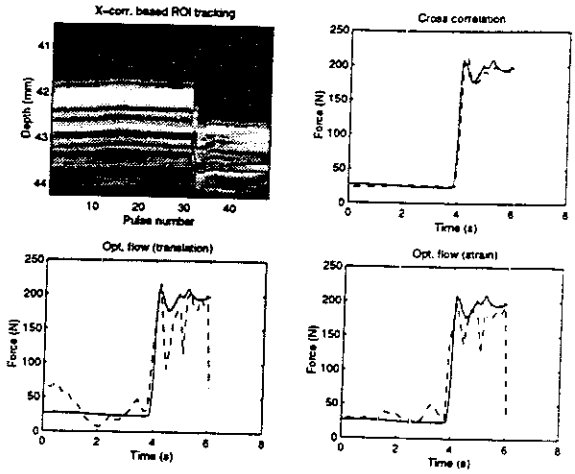

Figure 3: M-mode image with white lines indicating XC-based ROI tracking (upper left). The upper right figure and the lower left and right figure shows estimated force based on XC, OF translation and OF strain, respectively (dashed), and actual force recorded (solid).

RESULTS

Force-ultrasound echo correlation

Figure 2, bottom section, shows the force generated by the biceps muscle during one of the trials. Notice the deliberate step that takes place after approximately four seconds; the force changes from $25N$ to $200N$ which roughly corresponds to 10% and 90%, respectively, of the maximum voluntary contraction.

The top section of figure 2 shows a so-called ultrasound M-mode image of what is believed to be the humerus-soft tissue interface (the bright horizontal "line" in the image) The top and bottom parts of figure 2 were recorded simultaneously. Each column of the image represents one 500-sample RF echo signal; the RF signal is rectified and smoothed, and the resulting signal determines the pixel intensities of the corresponding image column. The M-mode image is generated by stacking subsequent columns horizontally; thus the vertical index of the image corresponds to tissue depth, while the horizontal index corresponds to ultrasound pulse number. The force step at $t = 4s$ is accompanied by an increase in the transversal dimension of the biceps muscle and thus in the distance between the ultrasound transducer and the humerus, which can be seen as a vertical shift in the M-mode image at $t = 4s$. It should be noted that this pictures an isometric contraction. Far more vertical translation is expected in the case of a nonisometric contraction because the muscle is then allowed to shorten and thus increase its transversal dimension even more.

Force estimation

The XC and OF based techniques for ROI tracking described in previous paragraphs were applied to the data set presented in figure 2. In figure 3 (upper left) the result of the XC based tracking is indicated by two lines that also show the depth and size of the ROI chosen. Since originally the force was sampled approximately 200 times faster than the M-mode columns, the recorded force signal was split at equidistant indices into the same number of intervals as the number of M-mode columns and the mean force in each interval was calculated. A second degree polynomial was fitted, in a least squares sense, to the resulting data sets to approximate the functional relationships between the force generated and the ultrasound based "tracks". The upper right and the two lower sections of figure 3 show the resulting force estimates.

For comparison, the root-mean-square (rms) of the recorded EMG was also calculated for a number of intervals corresponding to the number of M-mode columns. The raw force signal, the cross correlation based force estimate and the EMG rms are plotted in figure 4.

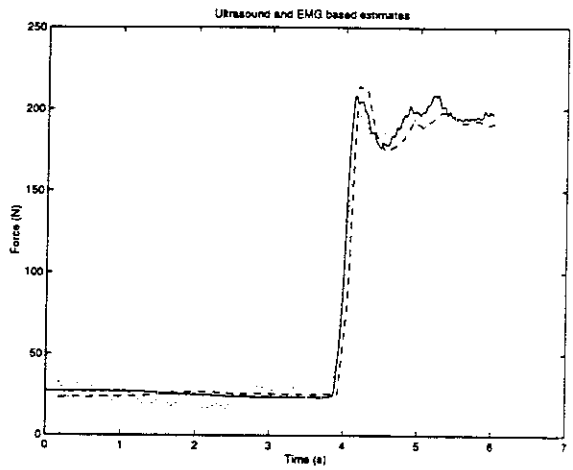

Figure 4: Raw force signal (solid), XC based (dashed) and EMG rms (dotted) estimates The EMG rms mean operation was performed over each interval corresponding to one M-mode column.

DISCUSSION

The results presented illustrate how, under ideal conditions, isometric contraction force estimates of relatively high quality can be derived from the ultrasound pulse-echo.

The OF based methods presented performed poorly compared to the XC based technique. In their paper, Bertrand et.al. suggest several methods for potential improvements, such as noise filtering of the ultrasound data and using pixels with large time and space derivatives only when evaluating of equation (2). We predict that the optical flow methods can be tuned to be more robust than the cross correlation one, because it takes into account not only the translational movement of the tissue but also the strain, which effectively makes the ultrasound signal decorrelate. The translational and the strain output of the OF method may also serve as input features to a more advanced system such as a pattern recognition module, and thus possibly further improve the contraction estimates. Several other techniques beside those mentioned here may be applied, including Doppler based methods for velocity estimation.

For the present technique to be applicable as a general method for muscle contraction estimation, its robustness with respect to probe position and orientation as well as long-term tissue changes must be addressed. Also, in the tracking operations presented one takes into account only the relative movement from one pulse-echo to the subsequent one, which eventually will cause the estimates to drift.

This study is meant to illustrate some of the possibilities that lie in the use of ultrasound measurements in connection with muscle related research and prosthesis control. One of the most important potentials is that of estimating contractions in multiple muscles from one single pulse-echo measurement. As opposed to surface EMG, the pulse-echo "looks" inside the tissue and can reveal information about multiple and deep-lying muscles. Moreover, ultrasound measurements are expected to be less sensitive to skin moisture variations than are EMG measurements; in fact, the more moisture, the better acoustic contact between the ultrasound probe and the skin. The problems related to the surface EMG being a function of a few motor units only and thus yielding poor force estimates probably are not relevant to ultrasound based measurements to the same extent. In the extreme case one may picture a two-dimensional ultrasound probe located inside a forearm prosthesis, yielding a complete cross-sectional image of the forearm and thus enabling force estimates to be calculated for any desired set of muscles.

Realizing that many practical and technical problems has to be overcome for these visions to become true, the authors feel that the potentials justify further research into this field.

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