Journal of Biomechanics xxx (2018) xxx–xxx

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Short communication

Perturbation velocity affects linearly estimated neuromechanical wrist joint properties

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ARTICLE INFO

Article history:
Accepted 1 April 2018
Available online xxxx

Keywords:
Wrist
Neuromechanics
Nonlinearity
Reflexes
Velocity

ABSTRACT

The dynamic behavior of the wrist joint is governed by nonlinear properties, yet applied mathematical models, used to describe the measured input-output (perturbation-response) relationship, are commonly linear. Consequently, the linearly estimated model parameters will depend on properties of the applied perturbation properties (such perturbation amplitude and velocity). We aimed to systematically address the effects of perturbation velocity on linearly estimated neuromechanical parameters.

Using a single axis manipulator ramp and hold perturbations were applied to the wrist joint. Effects of perturbation velocity (0.5, 1 and 3 rad/s) were investigated at multiple background torque levels (0, 0.5 and 1 N·m). With increasing perturbation velocity, estimated joint stiffness remained constant, while damping and reflex gain decreased. This variation in model parameters is dependent on background torque levels, i.e. muscle contraction.

These observations support the future development of nonlinear models that are capable of describing wrist joint behavior over a larger range of loading conditions, exceeding the restricted range of operation that is required for linearization.

1. Introduction

In (wrist) joint dynamics, the presence of nonlinear properties is evident. Nonlinearity can be illustrated by showing that linearly estimated dynamics of the wrist joint are dependent on the inputs applied, e.g. the degree of exerted forces (muscle contraction) or the magnitude or velocity of wrist rotation. For example, increasing muscle contraction results in increased joint stiffness due to the increase in muscle cross-bridge attachments (Zhang and Rymer, 1997; Hunter and Kearney, 1982; Gordon et al., 2000). Also, an increase in joint velocity is known to result in a reduced ability to produce muscle force, in line with the force-velocity relationship (Fenn and Marsh, 1935; Joyce et al., 1969; Westing et al., 1988; Hanten and Ramberg, 1988; Hill, 1938). Such nonlinear properties need to be considered in the (linear) identification of joint dynamics.

To gain an understanding of the etiology of clinical phenotypes and for the improvement of patient specific therapy it is important to access individual contributions of passive tissues, active muscles and the reflexive pathway to joint impedance (van der Krogt et al., 2012). Such measurement should be performed in tasks where movement disorders affect joint functioning. Any linear representation of dynamics (e.g., in terms of stiffness, damping) over such a range of conditions is expected to vary noticeably due to nonlinearity.

A full nonlinear model of the wrist joint is not yet available. This complicates individual assessment of reflexive and non-reflexive/ intrinsic (musculo-tendinous) contributions during functional movement. However, local linear analysis on small amplitude perturbations, in combination with system identification methods and simple linear model structures can be used to identify these components separately (Kearney et al., 1997; Van der Helm et al., 2002; Zhang and Rymer, 1997; Perreault et al., 2000).

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https://doi.org/10.1016/j.jbiomech.2018.04.007
0021-9290/

Using carefully controlled measurement conditions in which linearization is valid, we can systematically explore the nonlinear landscape of the complex wrist joint dynamics. This paper follows our work on effects of perturbation amplitude and contraction level on linearly estimated neuromechanical parameters (Klomp et al., 2014), where it was shown that a linear model can describe the joint dynamics well under similar measurement conditions.

The aim of this study is to systematically investigate the effects of perturbation *velocity* on linearly estimated neuromechanical parameters. This is a first step in trying to understand the effect of perturbation velocity on linearly estimated joint impedance, in terms of intrinsic and reflexive components, and will assist in future nonlinear modeling of the wrist joint.

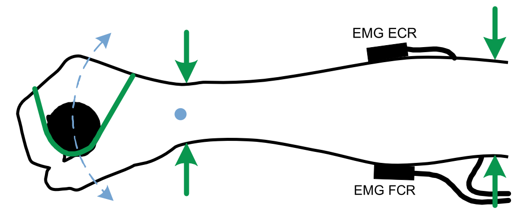

Fig. 1. General measurement setup. The green arrows indicate points of fixation, at the wrist and elbow, the green lines represent the straps used to fix the hand to the handle. The blue dot indicates the motor rotation axis and the blue dashed line the trajectory of handle rotation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

2. Methods

Apart from subjects and perturbation signals all methodology was consistent with Klomp et al. (2014). Please refer to that article for a more detailed description of general methodology and further review of the estimation procedure.

2.1. Subjects

Eight healthy subjects (33 ± 9 years, 3 female) participated in the study after providing a written informed consent. The study was approved by the local medical ethical committee.

2.2. Measurement protocol

Continuous angular perturbations of the wrist joint were applied by a single axis manipulator (see Fig. 1). The setup consisted of a vertically positioned electrical motor connected to a handle via a lever.

Joint angle was measured by an encoder (Stegmann SRS50, accuracy 0.011°) and torque was measured by a custom-made force transducer located in the lever (inaccuracy < 1% full scale). Muscle activity of mm. flexor and extensor carpi radialis (FCR and ECR, respectively) was measured using surface electromyography (EMG) electrodes (Delsys Bagnoli-4, inter electrode distance 10 mm), on-line band pass filtered (20–450 Hz), off-line rectified and integrated by low pass filtering (3th-order Butterworth) at 80 Hz. All signals were stored at 2 kHz sample frequency.

All tests lasted 25 s, of which the first 5 s were removed to reduce the influence of start-up behavior. Subjects were either asked to relax (do nothing), or to apply a constant flexion torque against the handle of the manipulator (i.e. lean against the handle) with an average torque of 0.5 or 1 N·m. While performing the flexion torque, the motor applied continuous angular ramp and hold perturbations (Fig. 2). These perturbations were applied with fixed velocity (0.5, 1 and 3 rad/s) and amplitude (0.15 rad peak to peak). Each of these 9 measurement conditions was applied 3 times, resulting in 27 tests, ordered randomly. For each measurement condition, two datasets were used for parameter estimation, and one for model validation. The highest velocity was chosen in line with previous work on perturbation amplitude (Klomp et al., 2014), and is known to yield reflexive activation. *Measured* joint angle is shown in Fig. 2. Filtered contact torque (second order low-pass, 2 Hz) was displayed on a monitor, showing a moving trace in time from the top to the bottom of the screen. Subjects were asked not to respond to the perturbation or excursions of the trace, yet slow corrections, to get the trace centered, were allowed. Subjects were given practice runs.

2.3. Modeling and analysis

A linear open loop neuromechanical model was used to describe the dynamical relationships between angle, torque and flexor EMG (Fig. 3). Musculo-tendinous viscoelasticity, as well as continuous muscle activation (voluntary and involuntary) was described by a linear inertia-spring-damper-model. Reflexive afferent feedback was modelled by a single reflex gain on joint velocity, in line with literature Kearney and Hunter (1988) and our previous findings (Klomp et al., 2013, 2014). As the velocity dependent activity (i.e., the dynamic phase) is only present during stretch, and not during shortening (Alfalahe et al., 1990; Edin and Vallbo, 1990; Matthews, 1981), velocity dependent reflexes were modelled unidirectional. The delayed feedback was compared to measured flexor EMG to improve the estimation of the reflex latency time and subsequently filtered by a second order filter representing the dynamics of an activating muscle. The filter was set by an estimated eigenfrequency, a fixed relative damping at 0.75 (Genadry et al., 1988), and an estimated gain for conversion from EMG to torque. The final reported reflex gain was estimated as two parameters placed in series, a reflex gain that related velocity with the EMG response (k_{v-E}) and the gain of the muscle activation dynamics filter (k_{E-T}), converting EMG amplitude to resultant joint torque ($k_v = k_{v-E} \times k_{E-T}$).

Estimated parameters were inertia (I), damping (B), stiffness (K), velocity dependent reflex gain (k_v), eigenfrequency of muscle activation dynamics (f_a) and the reflexive latency time (t_d). Search intervals are given in Fig. 4 and have been set the same as in our previous work on amplitude dependency (Klomp et al., 2014).

In ramp and hold perturbations, the contributions of I, B and K are temporally distributed on the start/stop, ramp and hold periods, respectively. Unidirectionality of the reflexive pathway, the neural delay and relatively slow muscle activation dynamics in the model differentiates the intrinsic and reflexive contributions.

To decrease computational effort of the optimization per test, data was averaged over consecutive ramps using a time window of −0.1 s with respect to ramp onset up to 0.2 s after the end of a ramp. Both directions, eccentric and concentric loading, were used in the optimization.

Validation Variance Accounted For (VAF*) was calculated to measure goodness of fit of the estimated model on a validation data set, which consequently also showed consistency (of the estimator and subject) over repetition.

Torque data in the [−0.1 to 0] s interval, prior to ramp onset, was reviewed to determine the consistency of the flexion torque in each subject. The average torque prior to each pair of consecutive ramps (flexion and extension) was calculated and the standard deviation (s.d.) of this average throughout each test was used to represent the consistency of the torque task within each of the measurement conditions.