We have developed a mathematical model based on the Hunt-Crossley's viscoelastic contact formulation for predicting the contact forces in the upper-body. The simulations were carried out in OpenSim software package and the simulations results were compared to experimentally recorded contact forces measured using a pressure algometer for assessing pressure pain sensitivity in the pelvic region 1.

We observed a very good agreement between the model prediction and algometer data. Our simulation revealed that by pressing down on the tissue both normal and frictional contact forces increase up to a point- ceiling effect. Moreover, viscoelastic properties of the examinee's tissue were associated with force; specifically, as the stiffness of the tissue declined both normal and frictional contact forces similarly declined albeit in a different way.

Once the contact force reaches a peak point (irrespective of the baseline stiffness of the tissue) additional pressure application by the examiner was associated with incremental decrease in both normal and frictional (wasted) contact force.

Citation


Introduction

Myofascial pain[2] is the most prevalent feature of chronic pelvic pain in men and women alike.

There is near universal agreement across disciplines that therapies targeting pelvic floor muscles are the most indispensable arsenal in treating patients with chronic pelvic pain [3-6]. Nonetheless, advances in mechanistic research, optimization and standardization of treatment regimen targeting pelvic muscles is critically hindered by fundamental deficit in objective measurement methodologies[1]; normative data is similarly absent in the field.

Scientific inquiry into etiology and mechanisms of pain in the pelvic region cannot be realized without quantifiable and reliable parameters describing pelvic floor function, parafunction, and clinically reliable and reproducible exam methodology. Spanning over two decades, finding an answer to these questions are and have been the focus of our team's research [1,7].

Without descriptive work on what is normal we are unable to quantify and compare therapeutic interventions. Moreover, we are unable to describe the impact of demographical variables on clinical measures and treatment outcomes alike.

We previously described the feasibility of obtaining quantitative measures of pressure pain sensitivity using a novel algometer and a standardized methodology.1 Building upon this initial
work, we then set out to define and develop additional tools for assessing number of other-as of yet unquantified-clinical variables which will be described in future studies. Our ultimate long-term vision is genesis of an in vitro mathematical model which in turn will enable inquiries into interplay among clinically relevant variables[8,9].

The specific focus of the present work is to investigate the dynamic interplay between examiner (biomechanical limitation of human finger with a focus on distribution of during prototypical exam) and examinee[10-13] (tissue laxity vs stiffness).

Clinical pain report is not a static data point. This is true across the board and not affected by assessment methodology (qualitative vs. quantitative). Quantitatively measured force in Newton (N) obtained by our pressure algometer (despite its validity and reproducibility) is not the same as temperature or weight. A wide range of intrinsic and extrinsic variables affect this number: tissue laxity or stiffness, patient related factors such as aversion and pain, positioning during exam (i.e. variation in positioning of the hip can lead to reflex contraction of pelvic muscles), inter-examiner's variation in exam and biomechanical characteristic of human finger. These are just a representative example of challenges encountered while standardizing clinical assessment [2,14,15].

Modeling techniques such as the one described in this study can help us overcome some of these seemingly insurmountable obstacles.

For our clinical colleagues, a prototypical patient scenario can best illustrate the clinical implication of this line of inquiry. A typical patient with a high tone tender muscle (hard viscoelastic property) is likely to experience pain at exceedingly lower levels (e.g.10 N where a 30 N force could otherwise have been tolerated if it were not for the presence of the pain). On the other hand, a patient with no pain and laxity (soft viscoelastic property) can similarly end up with the same number if not lower (e.g. 8 N). In this situation one has a valid data point from a device perspective but diabolically different clinical picture for the same number.

Unaccounted intrinsic [10-13] (e.g. muscle bulk and inter-individual variation) and extrinsic (e.g. childbirth, contraction of muscles during exam because of pain, clinician's approach) factors affect measured force. As exemplified by our second examinee, a person with loss of muscle mass, laxity and no pain can end up looking just like another person with pain who happens to have normal or hypertrophied muscle mass (hard viscoelastic property). Moreover, muscle tightness (hard viscoelastic property) can either be because of the muscle mass or aversive or involuntary tightening secondary to pain in the course the examination.

These are some of the challenges encountered using unadjusted raw datapoint in making clinical inferences. These issues are not unique to our field and have successfully been addressed in other scientific disciplines using biomedical and behavioral modeling[16] techniques.

Single digit examination- using examiner's index finger- is the corner stone of clinical assessment in medicine, especially in the pelvic region. While performance characteristics of human finger during clinical exam is unknown, significant advances in biomedical modeling of finger over the past decade now enables us to start this line of inquiry.

Fok[17] and Vigouroux[18] were among the first who utilized non-linear optimization method to minimize the tendon forces based on muscle physiologic cross-sectional area (PCSA) as the cost function. Others have applied direct solution method to solve the simultaneous equations of the joints in their model [19,20].

Thus, the aim of the present study was to assess feasibility of an in vitro model investigating the interface between examiner and examinee. Specifically we sought to understand the impact of viscoelastic examined tissue (laxity vs stiffness ) on distribution of the force along the kinetic chain of the examiner’s index finger: interphalangeal joints, metacarpal phalangeal joint, and upper
extremities’ muscle groups (extensor and flexor muscles)

Methods

Figure 1. Upper-Extremity right-side model

To assess the pressure distribution within different tissue types, it is critical to develop a predictive mathematical model capable of predicting the force distribution under various loading conditions. To this end, in this study, we have leveraged the Upper-Extremity model originally developed in the Stanford University to simulate the force distribution. The model originally was created in SIMM software package and we subsequently imported this model to OpenSim Figure 1 and utilized the Forward Dynamics approach to simulate the model. To better examine the kinematics of index
finger, we used Newton-Euler inverse dynamic approach. Three bony segments of the index finger were approximated by cylinders with circular cross sections Figure 2; geometrical and mass properties of the model segments are summarized in Table 1.

![Simplified index model with joints and segments](image)

Figure 2. Simplified index model with joints and segments

In order to express torque along the joint Po-Ling Kuo methodology was used[21]. This group have proposed that the equations for the torque of each joint would be expressed in terms of mass M, centrifuge R, coriolis C, gravity G, and external force F. For instance, from Figure 2, the equations for the second joint can be written in the following form.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Vol. (m$^3$)</th>
<th>Mass (kg)</th>
<th>$I_{yy}$ (m$^4$)</th>
<th>$I_{xx}=I_{zz}$ (m$^4$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>2proxph</td>
<td>8.3e-6</td>
<td>0.016</td>
<td>3.9e-7</td>
<td>4.1e-6</td>
</tr>
<tr>
<td>2midph</td>
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<td>0.0054</td>
<td>6.7e-8</td>
<td>2.4e-6</td>
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<tr>
<td>2distph</td>
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<td>0.0011</td>
<td>5e-9</td>
<td>1.6e-7</td>
</tr>
<tr>
<td>2proxph</td>
<td>2.0e-7</td>
<td>0.0004</td>
<td>8e-10</td>
<td>3.5e-8</td>
</tr>
</tbody>
</table>

Table 1. Geometrical and mass properties of the segments

\[ n_2 = M + R + C + G + F \] (1)

In the above equation, the terms of mass M, centrifuge R, coriolis C, gravity G, and external force F are derived from the following equations.
In the above equations and stands for the mass and length of the cylindrical bodies respectively Figure 2. Also, and represent the joint angular velocities and accelerations. In addition, the terms c and s have been utilized for cosine and sine, respectively. Furthermore, the moment of inertia for each body has been reported in Table 1. Moreover, to model the contact forces, since the index finger is covered by soft-viscoelastic layer, the dynamic model of the contact was modeled based on the Hunt-Crossley model[22], resulting simulations were compared against previously described experimental set up.

## Results

Measured contact force from the algometer device[1] was on average 18-25 N with maximum force ~ 30 N, consistent with our simulation.

Table 1 describes geometrical and mass properties of cylindrical representation of the distal, mid and proximal phalanx. Figure 1 show cases the 3D representation of the Hunt-Crossley model of the upper extremity, followed by the 2D simplified version of this model used for simulation of contact forces along the joints and bones of the hand (specifically index finger) in this study.

### Variation of force based on biomechanical parameters of index finger during exam
Figure 3. Normal contact forces variation with index finger’s distance from surface

Figure 3 and Figure 4 represent change in force during examination. The inflection of the tissue by the index finger is represented by different color with simulated distance from the surface (see colored legend). Distance away from the surface is represented in millimeter and is denoted as ‘negative number’ on the left-hand side of the figure along Y-axis, top of the figure along x-axis represents the contact surface and is denoted as ‘ground zero’ 0.0. Each colored graph within Figure 3 and Figure 4 represents the distribution of expected force during examination. Because contact and frictional forces are two complimentary derivatives of examiner applied force, they are best reviewed side by side for clinical context starting at midpoint (time = 0.5 seconds on x-axis) in Figure 3 and Figure 4.

Figure 4. Frictional contact forces variation with index finger’s distance from surface
Midpoint of the examination corresponds to t= 0.5 sec (total duration of exam = 1 second) on x-axis deflection depth of 40 mm (gray colored) on the y-axis. The upside-down gray curve in Figure 3, thus represents the distribution of force midway through the examination. Looking at Figure 3, one can see that midway during exam is when the highest portion of examiner applied force is transmitted to the underlying tissue with minimal to no frictional force as noted by absence of gray color distribution on Figure 4.

This mid-point in examination is the turning point after which a higher portion of the applied force is wasted as frictional contact force Figure 4 with progressive degradation of normal contact force Figure 3.

**Variation of force based on viscoelastic properties (tissue laxity vs stiffness) of the examinee’s tissue**

![Normal contact force VS. time](image)

**Figure 5. Normal contact force for different contact surfaces**

Figure 5 and Figure 6 represent the change in force during examination based on viscoelastic properties of the examinee’s tissue. In these graphs x-axis represents time with color legends in the box representing viscoelastic properties of the tissues: K1 in red represents a high resistant hard tissue, with K6 representing the opposite spectrum, low resistance, soft tissue (Figure 5 and Figure 6).
Resistance against applied force from tissue starts to build up slightly before halfway through the exam (t = 0.45 second) and it tends to increase differentially based on tissue stiffness. With hard tissue (K1 represented in red) one sees a sharp increase in both frictional and normal contact force followed by a similarly sharp decline. As the stiffness of the tissue decreases, flattening and spreading of the curve in both frictional and normal contact force is observed.

Simulation of force along the kinetic chain of the examiner’s hand

Figure 6. Frictional contact force for different contact surfaces

Figure 7. Flexor muscle forces for different contact surfaces
Lastly, we simulated the behavior of the flexor and extensor muscles (Figure 7 and Figure 8) -as they are in the same kinetic chain- as hand’s joint (interphalangeal and metacarpal phalangeal)[23] as hand (Figures 11–13).

Figure 7 and Figure 8 simulates the behavior of flexor and extensor muscles during examination. These muscle groups are engaged from the onset (stabilizers) and reach an equilibrium early on. In the latter half of the exam there is nominal degradation particularly along extensor muscle group (Figure 8).
Figure 9. Proximal interphalangeal joint flexion for different contact surfaces

Figure 10. Distal interphalangeal joint flexion for different contact surfaces

Figure 11. Metacarpal phalangeal joint flexion for different contact surfaces

Force along the hand’s joints were similarly affected in the latter half of the examination with greatest impact on distal interphalangeal joint (Figure 10) where stiffness of the tissue had a linear relationship to the amount of force at distal phalangeal joint. In other word, the stiffer the tissue
the more force is transmitted to this joint, more over this force progressively increases during examination. Unlike distal interphalangeal joint, however, tissue laxity had nominal effect on proximal interphalangeal joint (Figure 9). Metacarpal phalangeal joints showed the most robust and sharp decrease in force in the latter half of the examination across all simulated tissue types (Figure 11).

Discussion

Hunt-Crossley’s viscoelastic[22] contact model is in good accord agreement with experimental data. Our mathematical model was similarly consistent with our empirical observation that force incrementally increases with gradual deflection of tissue under the index finger until a point. Specifically, during exam, both normal and frictional contact forces (detail discussion below) increase until a point (ceiling effect) after which additional application pressure by the examinee leads to incremental decrease. This ceiling point is primarily determined by biomechanical characteristics of the human index finger, and static and dynamic viscoelastic properties of the tissue.

Biomechanical limits of human index finger during clinical exam is around 25-30 N. This prediction by the model is consistent with our observation and that of other investigators (personal communication, unpublished data). Moreover, viscoelastic properties of the tissue affect both normal and frictional force with greatest consequence at the distal interphalangeal joint. Stiffness of the tissue in our simulation led to progressive increases in load at this joint specially in the latter half of the examination.

Static viscoelastic properties of tissue are a composite measure. A given biological potential (lax vs stiff) is attenuated or potentiated by number of factors: a) demographical factors (e.g. age, gender), comorbid conditions outside pelvic (e.g. scoliosis, knee joint pain or low back pain indirectly affecting pelvic muscle), comorbid condition within pelvic region (e.g. Inflammatory disorders of bowel, bladder, and reproductive track), trauma (e.g. childbirth), iatrogenic (e.g. radiation, pelvic surgery). Absent ability to account for all these variables, we used a very strict enrollment criterion. Thus, our findings may not be generalizable to a larger clinical population which we intend to investigate in future validation cohort.

Dynamic viscoelastic properties are similarly a composite measure that occurs during the examination. Patient’s expectation, pain experience, and positioning for examination are some of the key determinants. While our study was not designed to account (adjust) for these variables we mitigated their influence by using a structured exam methodology (video links forthcoming).

To examine the clinical implication of normal vs frictional contact force we will use an analogy of a beam suspended by one end from a hinge while the other end is in contact with the ground. Human index finger as a dangling appendage (like a suspended beam) is an unstable mechanical system as it is supported by one end alone and its movement is orchestrated primarily by force directed at its ‘hinge’- metacarpal phalangeal (MCP) articulation. The fingertip in our analogy is like the free end of the beam in contact surface. If a force were to be applied at the hinge of the beam in our example, it will be distributed into two components direct force (normal contact force) will be the portion of the original force that is transmitted to the ground at its point of contact; this analogous to the forced applied by the fingertip. While the non-direct force is primarily wasted because of the instability of the mechanical system. Regardless of the “give” at the contact point with the ground (viscoelastic property), there is only so much force that gets transmitted to the ground; the rest is wasted. And depending on the amount of original force, subtracting the direct force, the remaining wasted force can lead to all of kind of movements by the beam: rotation of the beam along its axis, swaying side to side, grinding on the surface, or any combination of the above. The so- called wasted force in this analogy is ‘frictional contact force,’ in our model, whereas the intended force for the purpose of direct pressure application against a given surface is referred to as ‘normal
contact force.’

With this analogy in mind one can appreciate the challenges of placing perpendicular/direct pressure against a surface area at its point of contact by a bend finger. Like our beam, different scenarios will play out in the course of pressure application which is further complicated by the softness or hardness of contact surface (examinee’s tissue type). Regardless of the softness or hardness of the ground, however, there comes a point that additional force is wasted and transmitted to surrounding region along with a push back against the finger as a torque.

Unlike the example of beam, human finger benefits from tendons and muscles that provide ‘scaffolding’ and support limiting the amount of frictional force (wasted force) during its movement. Nonetheless, this principle holds true in that there is only so much force that can be applied with a bend finger during a prototypical pelvic exam. And that additional force using ancillary muscles in kinetic chain (i.e. forearm and arm) will be ineffective and wasted as frictional force. Like our beam, wasted force is transferred to surrounding tissue with inadvertent movement of the examiners hand in situ touching unintended region. Our simulation supports that notion that variation in patient pain report and poor inter-examiner correlation can in part be attributed to nuances of pressure application by the examiner[1].

Our modeling data highlights the necessity of standardized approach to clinical assessment. In our experience, the greatest barrier toward standardized approach to exam is errors of our assumptions about ‘routine-ness’ of our bread and butter physical exam and its validity and reliability. Knowing the biomechanical limits of index finger and awareness of our movement during the exam is perhaps the most important step toward uniform assessment. Use of tools such as algometer can potentially assist with training, education, and standardization of clinical assessment, nonetheless clinician’s awareness of these issues in our view are by far the most important factor in obtaining reliable and valid data points.

Having shown cased the feasibility of modeling index finger, we intend to formally examine the predictability and reliability of this model using our experimental data in conjunction with inclusion of key clinical variable such as age and parity in order to arrive at a mathematical correction factors for our algometer chief amongst them being the viscoelastic properties of the examinee.

Findings of our simulation on kinetic chain with differential loading of force on joints and muscles of arm and forearm may also be of potential interest for colleagues investigating occupational hazard, physical therapy and rehabilitation of upper extremity.

It is our vision that at the completion of this line of inquiry -a small step toward developing a standardized and reliable clinical biomarker- we will position the field of pelvic medicine to benefit from the advances made in greater world of medicine to arrive at individualized patient centered treatment algorithm and classification system.

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