

1 **An Optical Coherence Tomography (OCT)-based Air Jet**
2 **Indentation System for Measuring Mechanical Properties**
3 **of Soft Tissues**

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11 **Running Title: OCT-based Air Jet Indentation**

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18 **Abstract**

19 A novel noncontact indentation system with the combination of an air jet and optical
20 coherence tomography (OCT) was presented in this paper for the quantitative
21 measurement of mechanical properties of soft tissues. The key idea of this method is
22 to use a pressure-controlled air jet as an indenter to compress the soft tissue in a non-
23 contact way and utilize the OCT signals to extract the deformation induced. This
24 indentation system provides measurement and mapping of tissue elasticity for small
25 specimens with high scanning speed. Experiments were performed on 27 silicone
26 tissue-mimicking phantoms with different Young's moduli, which were also measured
27 by uniaxial compression tests. The regression coefficient of the indentation force to
28 the indentation depth (N/mm) was used as an indicator of the stiffness of tissue under
29 air jet indentation. Results showed that the stiffness coefficients measured by the
30 current system well correlated with the corresponding Young's moduli obtained by
31 conventional mechanical testing ($r = 0.89$, $p < 0.001$). Preliminary *in-vivo* tests also
32 showed that the change of soft tissue stiffness with and without the contraction of the
33 underlying muscles in the hand could be differentiated by the current measurement.
34 This system may have broad applications in tissue assessment and characterization
35 where alterations of mechanical properties are involved, in particular with the
36 potential of noncontact micro-indentation for tissues.

37

38 **Keywords:** indentation, ultrasound indentation, soft tissue, elasticity,
39 air-jet, optical coherence tomography

40 **1. INTRODUCTION**

41 In many tissue pathologies such as fibrosis, edema, and cancers in breast, liver and
42 prostate, change of mechanical properties [1-4] is a common phenomenon observed by
43 clinicians or patients themselves using the hand palpation. However, the hand palpation
44 method is qualitative, at most semi-quantitative (such as the scoring system from the
45 palpation impression), thus limiting its use in quantitative and objective studies. In the
46 last two decades, in order to find optimal modalities for disease diagnosis and tissue
47 assessment, more and more researchers are putting efforts in developing quantitative and
48 objective approaches in the field of elasticity measurement and imaging based on
49 ultrasound [5-15], MRI [16], or optical [17-22] measurements with the help of some kind
50 of mechanical disturbances caused to the tissues including compression, indentation,
51 suction, vibration, or acoustic radiation.

52 Indentation is currently one of the most frequently used techniques to measure the
53 biomechanical properties of soft tissues [23-24]. Indentation, due to a small contact area
54 with the tested objective, has the advantages of no necessity to excise the particular tissue,
55 which is almost not possible for a standard uniaxial compression test, thus allowing its
56 especial use for *in-vivo* applications. Traditional indentation system used a rigid indenter
57 to compress the tissue in order to characterize the mechanical properties of soft tissue
58 using the relationship between the force and deformation. However, usually no thickness
59 information could be obtained directly from the indentation test. Tissue thickness is an
60 important parameter for the diagnosis of some tissue pathologies such as cartilage
61 degeneration [25] and is also an important factor for the calculation of tissue stiffness
62 because this is included in the boundary condition of the theoretical analysis of the

63 indentation test [23]. The thickness was usually measured by extra post-test methods such
64 as the needle probe penetration [26]. To address this issue, Zheng and Mak [7] had
65 developed a portable ultrasound indentation system, which used the ultrasound transducer
66 itself as an indenter. The system is capable of measuring both the initial thickness and
67 stiffness. Due to its easy operation and a relatively compact profile fit for clinical
68 operations, this ultrasound indentation system has been successfully applied in the
69 assessment of a variety of tissues *in vivo*, including muscular tissues [27], residual limb
70 tissues [28], diabetic foot plantar tissues [29], neck tissue fibrosis induced by
71 radiotherapy [30-31], hypertrophic scar tissues [32-33], carpal tunnel ligament [34].
72 Similar ultrasound indentation has also been used for the assessment of articular cartilage
73 [35-36].

74 Typical ultrasound indentation uses an ultrasound transducer with the central
75 frequency of 2~10 MHz for which the resolution is quite limited for small specimens
76 such as the skin and articular cartilage. For transducers with a higher frequency the end is
77 normally concave for better energy focus and larger penetration depth. In this case the
78 transducer tip cannot be directly used as an indenter. Introduction of a bolster at the
79 transducer tip for a planar indentation surface may possibly induce a poor coupling of the
80 ultrasound signal and significantly attenuate the ultrasound signal, which severely affects
81 the ultrasound measurement [36]. Furthermore, a rigid indenter is limited in achieving a
82 fast scanning speed because of the requirement of point-wise measurement. To address
83 these issues, a water jet indentation system had been developed [37-38]. In this system,
84 the water jet serves as both the indenter and the coupling medium, thus significantly
85 improving the speed in a C-scan test, i.e. the imaging plane is perpendicular to the

86 ultrasound beam [38]. Phantom tests showed that this system was capable of measuring
87 the soft tissue elasticity quantitatively and reliably and also providing a fast C-scan
88 mapping of the tissue elasticity. It has been further applied for the study of cartilage
89 degeneration in a bovine patella model [39] and bone-tendon junction healing in a rabbit
90 model [40] *in vitro*. The non-contact fluid indentation method also have the advantage of
91 lowering the potential or risk of causing damage to the tested soft tissue, especially in
92 tissues such as in scars or wounds where the contact indenter may cause inflammatory
93 effects. Almost at the same time, another group reported a similar system using water jet
94 for the assessment of articular cartilage [19]. They used an optical technique to detect the
95 deformation applied by the water jet on the cartilage surface. Briefly, an optical beam
96 with a constant intensity was illuminated onto the cartilage surface, and reflecting light
97 was collected and its intensity was related to the local surface curvature, which was
98 caused by the compression of the water jet.

99 One inconvenience to use the water-jet indentation is the water will split all over
100 during the test. If it is used for tissues located inside body, the water or saline needs to be
101 removed by another instrument. Another important factor for the widespread applications
102 of prototype biomedical instrumentations to clinical situations is the potential of device
103 miniaturization [41]. Even though miniaturization of ultrasound transducers is possible,
104 such as those used for intra-vascular ultrasound imaging system, it is very expensive and
105 the resolution is inherently limited. The water jet optical measurement system [19] used
106 the intensity of the reflecting light to measure tissue deformation, but it is not accurate,
107 particularly when the deformation is large or the tissue surface has different colours and
108 textures.

109 In various optic methods, optical coherence Tomography (OCT) is a recent and fast
110 developing technique that has acquired more and more widespread use in biomedical
111 research [42]. The principle of OCT itself is analogous to that of the pulse-echo
112 ultrasound imaging. It collected backscattered signals from the optically scattering tissues
113 for the purpose of a cross-sectional imaging. The difference with respect to ultrasound is
114 that OCT uses the optical interferometric rather than absolute time of flight technique to
115 resolve the spatial information. Using an optical interferometric method, OCT has the
116 high resolution thanks to the small coherence length of the light sources. Axial and lateral
117 resolutions in a range of several microns can be achieved, thus making the optical biopsy
118 a unique and attractive characteristic of this technique [43-45]. Similar to the ultrasound
119 signals, the OCT can measure the thickness of tissue layers where normally at the
120 interfaces the refractive index is abruptly changed. For example, one of the most
121 important applications of OCT is pachymetry, i.e., to measure the corneal thickness *in*
122 *vivo* for the use of diagnosis or surgery guide [46-47]. OCT-based elasticity imaging for
123 tissues has also been widely investigated using contact compression [18,20-21]. In
124 comparison with ultrasound elastography, OCT-based technique can provide higher
125 resolutions. However, these elasticity imaging techniques can normally provide contrast
126 of local strains but not an absolute value of tissue elasticity.

127 The integration of an OCT probe with the air jet indentation was realized in the
128 current study to develop a novel system to measure the mechanical properties of soft
129 tissues. The construction of the system was first described in the next section, then
130 experimental and data analysis methods on phantoms and *in vivo* hand soft tissues, and
131 the corresponding results were presented to demonstrate the utility of the current system,

132 and finally further issues related to the limitations, further improvement and applications
133 of the current system in biomedical engineering were discussed before the study was
134 concluded.

135

136 **2. METHODS**

137 *2.1. System Setup*

138 The schematic of the OCT-based air jet indentation system including the core part
139 and the data collection part is shown in Figure 1 and Figure 2, respectively. The fiber-
140 based OCT probe was modified to allow the installation of an air jet bubbler. The OCT
141 system (Developed by Lab of Optical Imaging and Sensing, Graduate School at
142 Shenzhen, Tsinghua University, China) has a super luminescent diode (SLD) light source
143 (DenseLight, DL-CS3055A, Singapore) with a central wavelength of 1310 nm, a nominal
144 -3 dB spectral bandwidth of 50 nm and a nominal output power of 5 mW. The axial
145 resolution is 18 μm and the imaging depth is approximately 2~3 mm in turbid high
146 scattering tissues. The OCT probe was fixed in this study and the laser beam focuses
147 vertically at around 5 mm under the lower surface of the bubbler. For convenience of the
148 detection, a visible red light beam was used with the invisible infrared beam to guide the
149 detection point. A pipeline with maximally constant air pressure was connected to the
150 system to provide the air jet for the indentation. To make the air jet more uniform, a tube
151 with an orifice diameter of 2 mm and a length of 5 mm was installed at the tip of bubbler
152 to guide the air jet before it was pushed into free space. A calibrated pressure sensor
153 (PMP 1400, GE Druck, Leicester, England) with a measurement range of 1 bar (10^5 Pa)
154 was installed before the bubbler to measure the pressure in the air pipe. A mechanical

155 valve was installed before the pressure sensor to adjust the pressure of the air jet
156 continuously. A transparent plate was installed at the top of the bubbler to seal the
157 pressurized air from the OCT components but let the laser beam to pass through.

158 A PC was used to control the operation of the system. A data acquisition card (DAQ,
159 PCI-6251, National Instruments, Austin, TX, USA) was used through
160 intercommunication to both control the main unit of OCT and collect the optical signals.
161 Another DAQ card (PCI-6024E, National Instruments, Austin, TX, USA) was used to
162 collect the signal from the pressure sensor. A custom-written program was developed in
163 Microsoft VC++ for the signal synchronization and data collection (Figure 3). During the
164 indentation process, the signals from the OCT and the pressure sensor were synchronized,
165 sampled, displayed in real-time and saved for off-line processing by the program. In this
166 study, the OCT scanning was continuously performed at a single location to track the
167 surface displacement, which was assumed to be equivalent to the deformation of the
168 tested specimen. The digitized A-scan signal was acquired at a rate of approximately 3.1
169 Hz and the pressure signal was also sampled at this rate in synchronization with the A-
170 scan signal. Each A-scan signal contained 7500 points of effective digital data for
171 analysis. By using the two surfaces of a glass slide with a standard thickness for distance
172 calibration, we obtained an equivalence of $0.43 \mu\text{m}$ /per point for each A-line of the signal
173 in air after calibration considering the difference of the refractive indices in glass and air.
174 The displacement of the surface was extracted by applying a cross-correlation algorithm
175 to the A-line signals, which was used to seek the most similar part to a pre-selected
176 region of interest (ROI) in the A-scan signals recorded during the indentation process.
177 The air-specimen interface was used as the pre-selected ROI in this study. In order to

178 reduce the effect of signal phase change during indentation and assure a better tracking,
179 amplitude signal obtained by applying a Hilbert transform to the original optical signal
180 was employed for the tracking. The same software using a different control panel was
181 used to conduct the algorithms of cross-correlation for the extraction of the deformation
182 (Figure 3).

183 2.2. *Phantoms and In-Vivo Tests*

184 We tested the system on 27 uniform tissue-mimicking silicone phantoms with
185 different stiffness. They were made of three kinds of silicone materials with low viscosity:
186 Rhodia RTV 573 (Rhodia Inc., CN7500, Cranbury, NJ, USA), Wacker M4648 and
187 M4640 (Wacker Chemicals Hong Kong Ltd., HK, China). For each type of silicone, there
188 were two parts A & B before mixing to make the phantom. Different proportions of A &
189 B were mixed to make phantoms with different stiffness. Here we chose three different
190 silicone materials to make phantoms covering a wide range of stiffness because phantom
191 stiffness was not only dependent on the mixing ratio of the two parts but also on the
192 silicone type. Three categories of phantom dimensions with a surface area of 10×10
193 mm^2 and a thickness of 5, 10 and 15 mm were fabricated [37]. All the experiments were
194 conducted on the $10 \times 10 \text{ mm}^2$ surface but with different phantom initial thickness (5, 10
195 or 15 mm). During the air jet indentation, the phantom was fixed at edges by four screws
196 and the air jet was exerted at the center of the surface. For each phantom, the axial
197 distance of the phantom surface to the bubbler bottom was adjusted to be approximately 5
198 mm based on the optical signal detected from the surface. If the specimen was too close
199 to the bubbler, it would be attracted toward the bubbler by a force induced by the high
200 velocity of the scattered air jet and the low pressure incurred around. Before each

201 indentation, the phantom was compressed and relaxed for several times to obtain a rigid
202 fixation on the platform. Then the air jet indentation was exerted at a pressure changing
203 rate of approximately 10 kPa/s and between 0 and 100 kPa, which corresponded to a
204 maximum displacement of about 0.37 mm for the softest phantom. The maximum
205 indentation depth was less than 5% for the phantom test. A typical test, including 1 to 2
206 cycles of indentation, was finished in approximately 30 s. For the purpose of comparison,
207 indentation tests with a rigid steel indenter (called standard indentation) and standard
208 compression tests were also performed using a standard mechanical testing machine
209 (Instron 5569, Norwood, MA, USA). For these two mechanical tests, the maximum
210 indentation or compression depth was about 10% of the initial thickness (only the data
211 within 3% deformation were used for parameter calculation) and the indentation speed
212 was controlled to be 4 mm/min, which was similar to that of the air jet indentation.

213 In order to demonstrate the biomechanical applications of the system, a preliminary
214 test was performed on soft tissues of the hand *in vivo* to differentiate the contraction state
215 of the underlying muscle layer involved. Ten subjects (7 males, 3 females) without any
216 lesions in the hand and with a mean age of 27.8 ± 2.9 years (minimum: 23; maximum 32)
217 were recruited in this test. A test site near the basal joint of the dorsal hand including the
218 muscles of the first interosseous was selected for experiment (Figure 4). The subjects
219 were asked to be seated in a natural posture with their hand placed on the platform. Two
220 states of the muscle, i.e. relaxation and contraction, were produced by natural extension
221 and forced adduction of the thumb (Figure 4). A deformation of approximately 1.5 mm
222 was applied on the tissues using the air jet. The corresponding indentation deformation
223 and forces were then collected and used for the calculation of the stiffness parameter.

224 **2.3. Data Analysis Method**

225 In the current study, the air jet indentation was assumed to be similar to the
226 indentation with a rigid contact, by hypothesizing that the air pressure measured in the
227 pipe was linearly proportional to the force induced on the specimen. The proportionality
228 of the fluid pressure and the indentation force was earlier validated for the water jet case
229 [37]. For the phantom tests, a stiffness coefficient k_{aj} (N/mm) regarding the regression
230 ratio of the air jet indentation force (N) and deformation (mm) of the tested specimen was
231 used to represent the stiffness of the phantom. This parameter was used here because with
232 the assumption of a linear elasticity, a constant Poisson's ratio and a small aspect ratio
233 (indenter radius/initial thickness a/h , in this study it was all less than 0.2), the
234 indentation formula was [23]:

235
$$E = (1 - \nu^2) / (2a\kappa(\nu, a/h)) \cdot F/d \quad (1)$$

236 where E the Young's modulus of the tissue, a the radius of the indenter, h the initial
237 thickness of the tissue, ν the Poisson's ratio of the tissue, κ a scaling factor related to ν
238 and a/h , F the indentation force and d the indentation depth. For the elastic material,
239 Equation 1 is simplified to:

240
$$E = (1 - \nu^2) / (2a) \cdot F/d \quad (2)$$

241 with κ approaching to one in Equation 1 [23,29,48]. The indentation force was
242 computed by multiplying the measured air pressure with the area of the bubbler orifice
243 ($\phi = 2a = 2 \text{ mm}$). The corresponding stiffness coefficient obtained by the standard
244 indentation was indicated by k_{std} . For the standard compression test, the Young's
245 modulus (E) of phantom could be obtained after computing the corresponding stress and
246 strain. We assumed a homogeneity of the compressive properties of the phantom so one

247 value of Young's modulus would represent all at each specific point. The deformation
248 ratio (deformation/initial thickness) or strain was constrained to be within 3% in all the
249 calculation in order to obtain a coefficient within the linear elasticity region of
250 compression of the phantoms [49]. Due to a low viscosity of the silicone phantoms, data
251 of both the loading and unloading processes were utilized for the calculation of the
252 stiffness coefficient. Pearson correlation coefficient was used to indicate the relationship
253 among the measured parameters from the air jet indentation, the standard indentation and
254 compression test.

255 For the *in-vivo* test, a corresponding stiffness coefficient k (N/mm) regarding the
256 force/deformation ratio was calculated from the test and the average value of three
257 repeated tests was used for each muscle contraction state. It should also be noted the
258 mechanical properties measured from the hand soft tissue were characteristic of the
259 whole tissue layer which might include the layers of skin, fat and muscles, but not that of
260 a separate single layer. When the whole tissue layer undergoes some pathological or
261 physiological changes such as the muscle contraction in the current study, the whole
262 tissue will behave differently. Therefore, the measurement of overall tissue properties is
263 also meaningful for the detection of tissue state under these conditions. A paired t -test
264 was used to compare the change of the stiffness coefficient k for the soft tissues with and
265 without muscle contraction. All the statistical analyses were performed in SPSS 14.0
266 (SPSS, Chicago, IL, USA). In all the statistical tests, $p < 0.05$ was used to indicate a
267 significant correlation or a significant difference of the mean between two measures.

268

269 **3. RESULTS**

270 The results of a representative indentation test on the phantom are shown in Figure 5.
271 Using the cross-correlation algorithm, the displacement could be successfully extracted
272 out during the indentation in the phantom tests. Some noise with the extracted
273 deformation was found during the processing, for which the reason was discussed in
274 detail in the next section. A high correlation coefficient, normally larger than 0.95, was
275 observed for the indentation force and deformation. A reliability test of 10 times of
276 repeated experiments on one phantom showed that k_{aj} was 1.108 ± 0.035 N/mm (3.2%
277 for the coefficient of variation), which showed the test was highly reliable. Correlation
278 tests among k_{aj} , k_{std} and E are shown in Figure 6. A high correlation of $r = 0.88$
279 ($p < 0.001$) was found for k_{aj} and k_{std} , which showed that the results of the two tests
280 were highly comparable. A larger value of k_{aj} was consistently observed from the air jet
281 indentation system than k_{std} , which might be attributed to the fact that a larger pressure
282 was measured inside the air pipe than at the surface of the specimen where the air jet
283 induced the deformation. The compression test showed that the stiffness of the phantoms
284 was 558 ± 124 kPa with a range of $328 \sim 818$ kPa. A comparison between k_{aj} and E
285 also showed a high correlation ($r = 0.89$, $p < 0.001$), which indicated the current system
286 could be used as a new approach to measure the mechanical properties of soft tissues.

287 *In-vivo* test showed that the stiffness coefficient was 0.059 ± 0.031 N/mm and 0.150
288 ± 0.059 N/mm with the muscle in the relaxation and contraction states, respectively. A
289 paired *t*-test showed that the soft tissue was significantly stiffer in the state of muscle

290 contraction ($p < 0.001$). Therefore, the current system was capable of differentiating the
291 state of the muscle contraction.

292

293 **4. DISCUSSION**

294 An air jet indentation system utilizing the optical signals from the OCT system was
295 developed and tested to demonstrate its capability in measuring the mechanical properties
296 of soft tissues. Preliminary results on silicone phantoms and *in vivo* soft tissues of the
297 hand showed that it was feasible to use the air jet indentation system to quantitatively
298 measure the stiffness and further to detect the change of stiffness that might be involved
299 in various tissue pathologies. The air jet was thought to be very convenient for use,
300 because no extra mechanism was needed to collect the splitting medium, as in the water
301 jet case. The incorporation of the OCT signal enabled the detection of deformation
302 (displacement) as small as less than 1 μm , and provided the potential for miniaturization
303 for use in applications such as endoscopy. Compared to the rigid contact indentation
304 method, this novel air jet indentation system also provided a potential for testing in fine
305 tissues and a fast scanning to map the deformation distribution in soft tissues.

306 In the current study, OCT signals were introduced to extract the deformation of the
307 indented specimen. The deformation was obtained by assuming that the platform
308 underlying the specimen was fixed and then it was equivalent to track the surface
309 movement. In real situations, the deformation as well as the thickness of the tissue can be
310 measured directly from the OCT signal, provided that the whole tissue layer can be
311 penetrated by the optical beam. This is also the advantage of using OCT for the air-jet
312 measurement system in comparison with other optical methods for surface displacement

313 measurement. In the current study, the measured stiffness coefficient represents the
314 mechanical properties of the whole tissue layer. During the data processing, it was found
315 that the optical signals were quite sensitive to the movement of the surface of the
316 specimen, even in the phantom test where the phantoms were firmly fixed. This might be
317 caused by the surface roughness of the skin and the orientation change of the optical
318 beam with non-vertical incidence during indentation. This sensitivity had brought some
319 noise to the deformation in tracking. When necessary, the moving average was used in
320 the current study to reduce the effect of noise. The movement sensitivity was one of the
321 big differences noted by us between the OCT and ultrasound signals. We had not
322 encountered such a problem previously in processing the ultrasound signal. This might be
323 due to the uncertain change of the phases in the OCT signals with respect to the surface
324 signal. This sensitivity might induce some tracking challenges in experiments,
325 particularly tests on tissues *in vivo*, where the tissue self-motion during indentation is
326 inevitable such as that caused by the respiration and heartbeat. Therefore in the current
327 design of soft tissue test *in vivo*, a roughly constant indentation depth was produced by
328 the air jet and then we measured the corresponding force in order to calculate the stiffness
329 coefficient. Another issue for using the cross-correlation algorithm was the decorrelation
330 caused by a large displacement. However, we thought the effect of decorrelation might
331 not be so significant in the current study due to two main reasons: one was that the
332 envelop signal was obtained to exclude or reduce the effect of phase change in tracking;
333 the other was that we only detected the abrupt change of the surface signal, which was
334 relatively easier to be tracked by choosing the ascending part of the signal at the surface.
335 The sensitivity of signal shape and signal to noise ratio to the movement tracking of the

336 specimen surface and approaches for lowering this sensitivity for a more reliable tracking
337 need to be further investigated.

338 The axial resolution of the OCT system used in this study was $18\ \mu\text{m}$, which was the
339 limit for the system to differentiate signals generated by two neighboring interfaces.
340 Since in the current study we were interested in the movement of a distinguished
341 interface, i.e. phantom surface or skin surface, we were able to achieve a much higher
342 displacement resolution beyond the axial resolution, given that the decorrelation of the
343 signal was small. The displacement resolution for the distinguished surface was mainly
344 determined by the number of data points for a certain period of OCT signal, which
345 represented a certain distance in the medium. In this study, we collected 7500 data points
346 for a distance of approximately $3.22\ \text{mm}$ (this value depends on the refractive index of
347 the medium). Therefore, the displacement resolution of the interface movement was
348 approximately $0.43\ \mu\text{m}$. Such a displacement resolution for interface tracking can be
349 achieved even when the OCT signals generated by two neighboring interfaces was
350 overlapped, i.e. the distance between the two interfaces is smaller than the axial
351 resolution of the measurement system, given that the two interfaces have no relative
352 motion during the movement. However, the displacement resolution discussed above can
353 not be applied to the movements of interfaces with the distance smaller than the axial
354 resolution and involving relative motions, as this will cause decorrelation to the OCT
355 signals. Under this condition, we are not able to tell which interface contributes to the
356 displacement of the signal and the displacement resolution will be significantly reduced.
357 The above discussion can not only be applied to the movement of the surface interfaces
358 but also those inside the media. Apparently, the displacement resolution of $0.43\ \mu\text{m}$ for

359 the interface movement cannot be applied to the movement of scatters inside the medium,
360 as the distances among scatters are very small and a smaller relative motion among them
361 will cause large decorrelation to the signal. That is why the displacement resolution for
362 ultrasound elastography [5], which aims to map tissue displacement or strain but not to
363 measure the displacement of a certain interface, is normally limited by the axial
364 resolution of the imaging system. In addition to signal decorrelation, cross-correlation
365 tracking for signals can also be affected by the signal to noise ratio and the sampling rate
366 of the signals. To measure tissue motion more accurately, it is worthwhile to
367 systematically investigate the displacement resolution of the proposed OCT air jet system
368 for the tissue interface and scatters inside the tissue in future studies.

369 A single point of displacement was measured in the current study using the A-line
370 OCT signal to obtain the deformation of the whole specimen layer. In further studies,
371 cross-sectional scanning, as used in B-mode ultrasound imaging, can be used to further
372 study the problem such as the deformation profile under the air jet, as long as the lateral
373 scanning can cover the size of the bubbler orifice. The frame rate of OCT A-line used in
374 this study was relatively low. Though it did not affect the measurement for the phantoms
375 and the carefully selected region of the hand, it should be significantly improved when a
376 cross-sectional scanning is required. Utilizing the cross-sectional imaging, the
377 elastography, which has been studied extensively in ultrasound and MR imaging field
378 [5,16], can also be performed, provided the air jet induces the required deformation to the
379 specimen [18]. One of the advantages of OCT compared to ultrasound is that the OCT is
380 easier for minimization due to the fast development of optical technologies and devices.
381 OCT probes for the endoscopic use in human mucosa had already been reported in the

382 literature [50-51]. Therefore, another aspect of further development is the minimization
383 of the air jet indentation system for portable or endoscopic uses [50-52]. In such cases,
384 the probe can be portable for flexible use in various body sites or small enough to be
385 inserted in endoscopy for internal tissue detection.

386 A hypothesis of the current study was that the air jet indentation was comparable to
387 the standard indentation with a rigid indenter and then the stiffness of specimen was
388 simplified in an index coefficient of the force/deformation ratio. The high correlation of
389 the stiffness coefficients between the air jet indentation and standard indentation really
390 showed that these two tests were similar. The high correlation between the stiffness
391 coefficient and the Young's modulus showed in a further step that the stiffness
392 coefficient could be used as a quantitative measure of the specimen elasticity. Thus the
393 developed OCT-based air jet indentation system can be further applied to a host of body
394 tissues for clinical diagnoses, such as skin cancer, burn status, corneal condition, blood
395 vessel stiffening and articular cartilage degeneration, where the change of tissue elasticity
396 is obviously involved from clinical observations.

397 However, intrinsic mechanical properties such as the Young's modulus are more
398 preferable for widespread applications because this will further enable the inter-lab or
399 inter-hospital comparisons of respective studies. Therefore, further investigations are
400 needed to extract the intrinsic mechanical properties from the air jet indentation test. This
401 is thought to be complicated due to two main reasons: complicated mechanical behaviors
402 of the soft tissues under rigid indentation, and the complicated interactions between the
403 air jet and the soft tissues in air jet indentation. The first issue was addressed previously
404 in a lot of studies [23,48,53-55]. In linear elasticity theory, the factors that affect the

405 extractions of the elastic modulus include the Poisson's ratio, indenter shape,
406 force/indentation ratio and the initial thickness. When the theory is applied to real tissues
407 *in vivo*, careful protocols should be designed to account for the extra effects from
408 viscoelasticity, nonlinearity, non-homogeneity and anisotropy of the tissue properties. In
409 this study, homogeneous and isotropic silicone phantoms were fabricated so as to
410 simplify the mechanical behavior of the specimen. The stiffness coefficient or Young's
411 modulus was obtained with the constraint of 3% of the deformation/thickness ratio or
412 strain, which was thought to be a linear elasticity region for most soft tissues [49]. The
413 viscosity was neglected in the study because it was not so obviously observed for the
414 silicone phantoms fabricated in this study.

415 On the other hand, the interactions between the air jet and soft tissues are much less
416 studied because the air jet indentation for the measurement of tissue elasticity was a novel
417 approach in biomedical engineering field. The waterjetting, similar to the air jet, is a
418 technique which has found widespread applications in a variety of industries [56].
419 However, the previous analysis on waterjetting could not be directly applied to our
420 studies because most of the industrial applications focus on material cutting and cleaning,
421 where the water pressure is much higher than that used in the current study. The
422 difference between the rigid indention and air jet indentation is that the tissue under the
423 rigid indenter is uniformly compressed with a planar surface in the former case while this
424 may not be the truth for the air jet one. It is expected that the deformation profile as well
425 as the pressure distribution in the interaction surface keeps changing as the change of the
426 air jet pressure. Finite element analysis, as used in previously indentation studies [57-60],
427 may be incorporated in further studies for the analysis of the air jet indentation to

428 investigate the interactions between the air jet and tissues and the effects of the variations
429 including the Poisson's ratio, the air jet radius, the distance between the bubbler tip and
430 the tissue surface, and the tissue thickness.

431 **5. Conclusion**

432 A novel OCT-based air jet indentation system was developed in the current study.
433 OCT can be used to detect tissue deformation as small as submicrons in a noncontact way.
434 The high correlation of the stiffness measured by the current system with that obtained by
435 the conventional and standard methods indicated that mechanical test with the OCT-
436 based air jet indentation was feasible. The capacity of the system to detect the
437 biomechanical changes in soft tissues had been demonstrated by phantom study and
438 preliminary *in-vivo* test. Further improvements are required to include lateral scanning
439 functions so as to map the tissue elasticity and to enhance the stability of OCT signal
440 from skin surface for *in-vivo* applications where the motion artifacts cannot be avoided.
441 The combination of the proposed air jet technique into the OCT elastography methods
442 recently reported in the literature [18,20-21,61-62] may be able to provide noncontact
443 OCT elasticity imaging for tissues. Testing on more tissues to demonstrate the potential
444 of this system for widespread applications in biomedical engineering is under planning.

445

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619 **Figure captions:**

620 Figure 1. (a) Schematic of the OCT-based air jet indentation system; (b) A picture of the
621 real system.

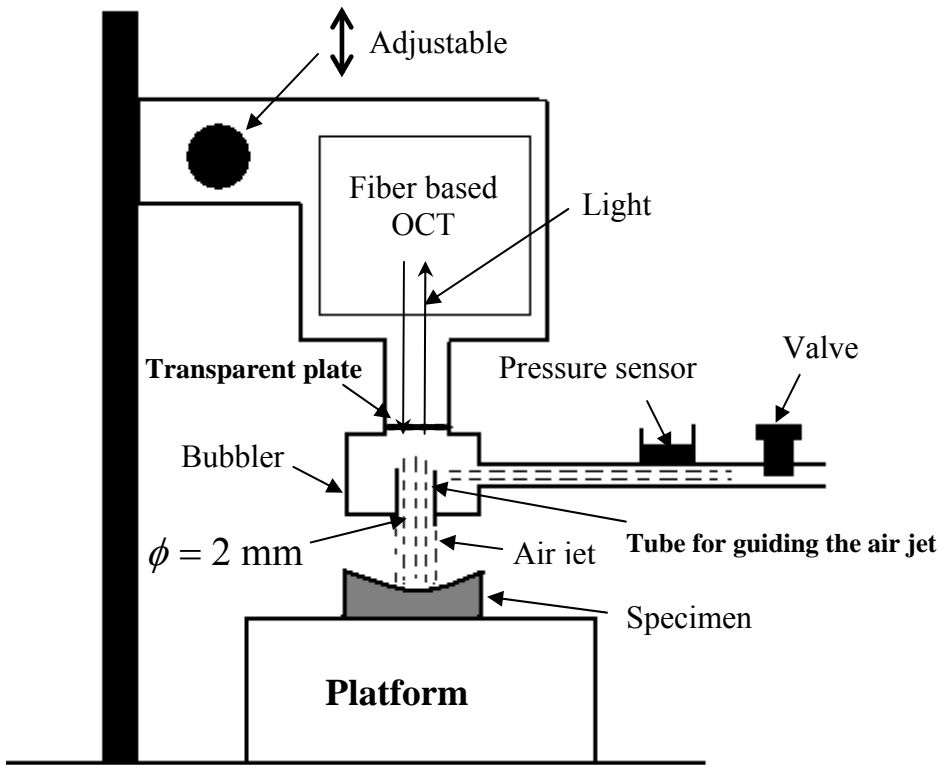
622 Figure 2. The diagram of the air jet indentation and data collection modules of the air jet
623 indentation system.

624 Figure 3. The custom-designed software interface for the real-time data acquisition and
625 post-acquisition off-line processing such as the displacement extraction. The left of the
626 window shows the control panels for data acquisition and post-acquisition processing.
627 The measured pressure and extracted deformation as well as the OCT signal are
628 displayed on the right.

629 Figure 4. *In vivo* experiment on the hand soft tissues. (a) Soft tissues indented without
630 muscle contraction; (b) Soft tissues with muscle contraction.

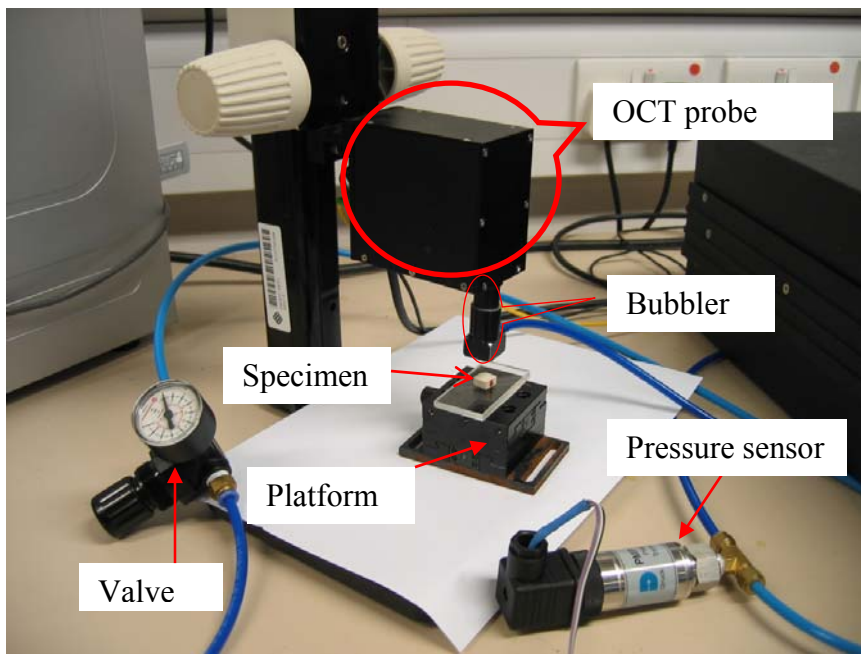
631 Figure 5. Representative indentation curves on one phantom. (a) Force-deformation
632 curves obtained during loading and unloading cycles; (b) The relationship between the
633 force and deformation of the phantom.

634 Figure 6. (a) Correlation of stiffness coefficients measured by the air jet and standard
635 indentation test; (b) correlation of stiffness coefficient measured by the air jet indentation
636 test and the Young's modulus measured by standard compression test.



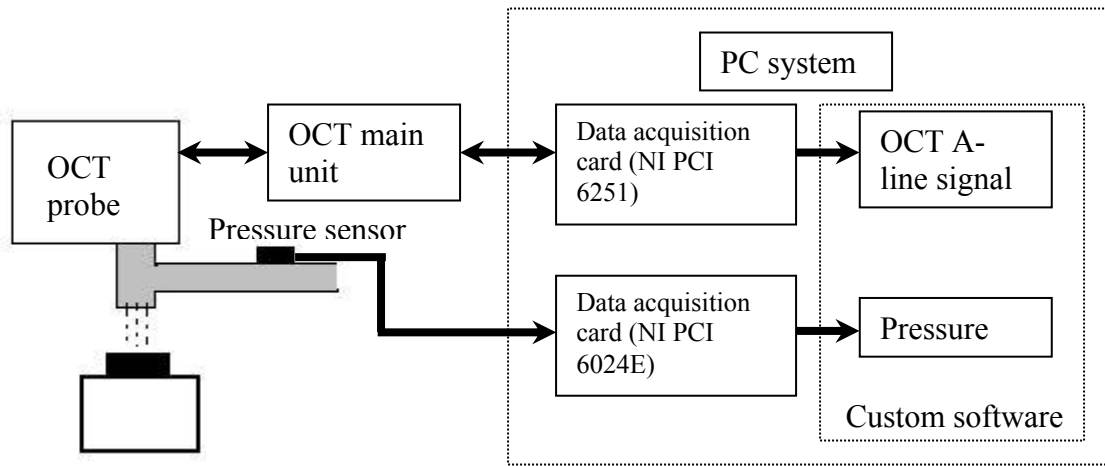
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638 Fig. 1 (a)



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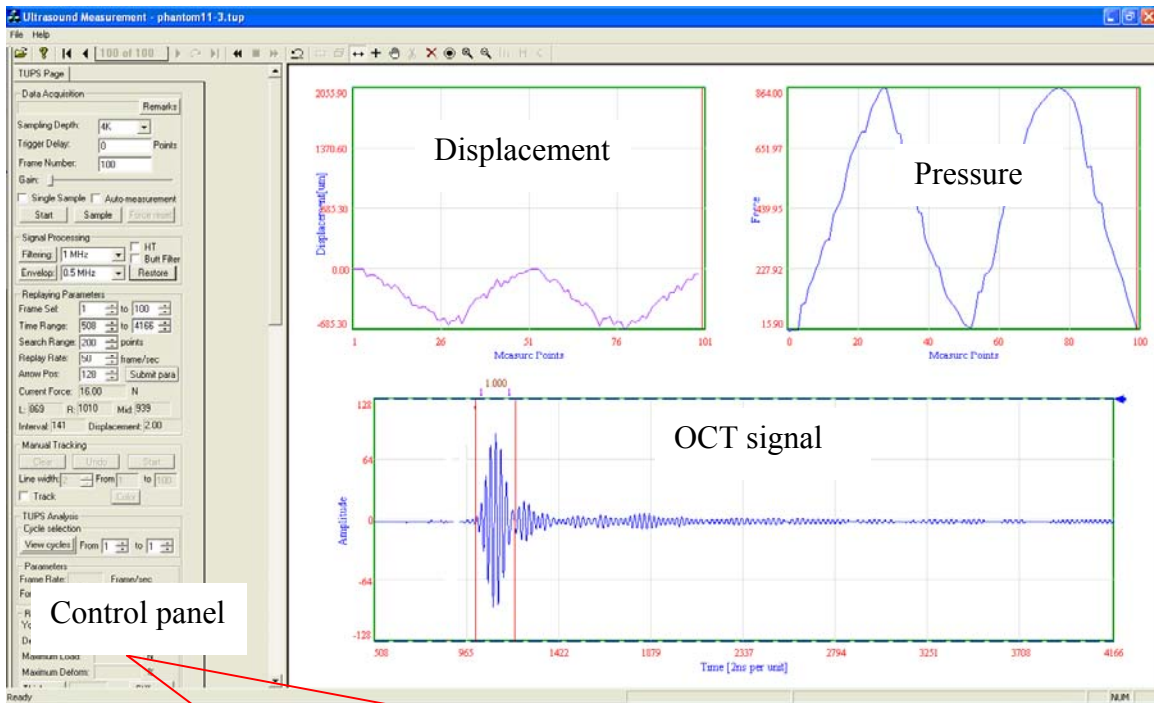
640 Fig. 1 (b)



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642

643 Fig. 2



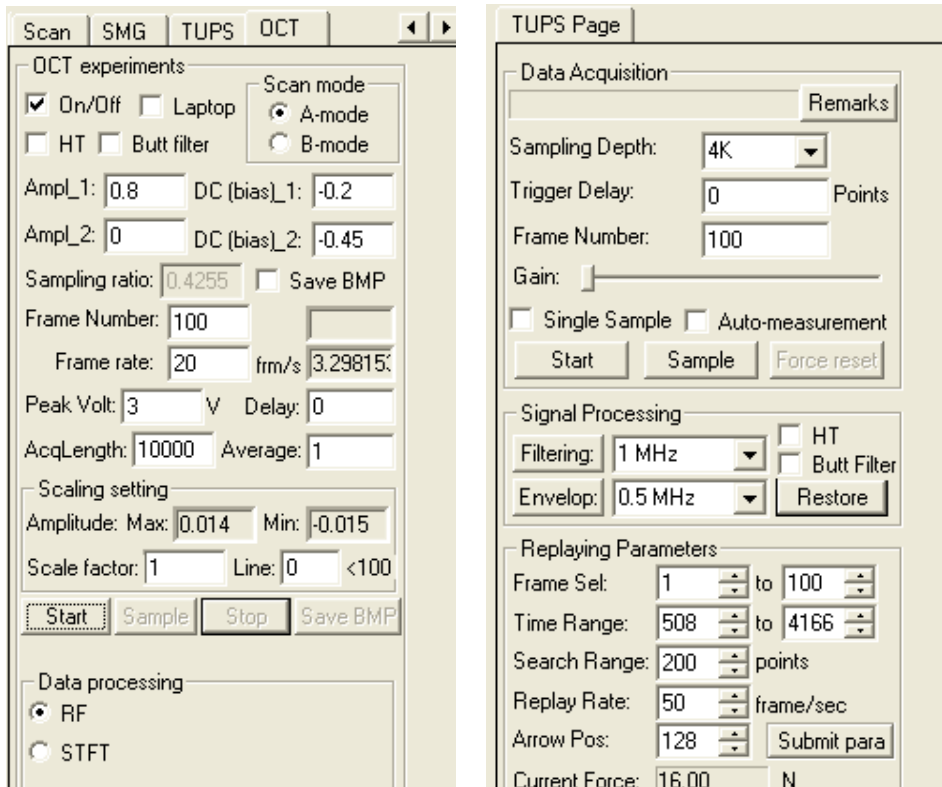
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Control panel

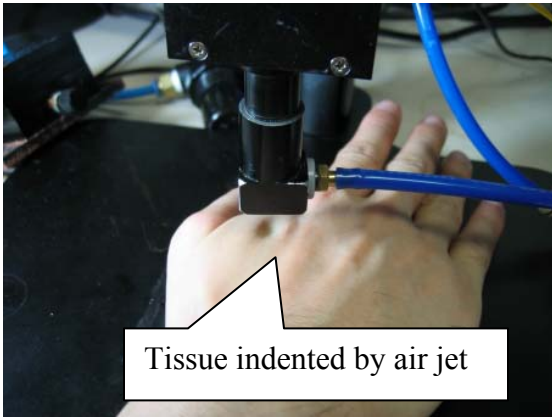
Used for acquisition

Used for post-acquisition processing



646

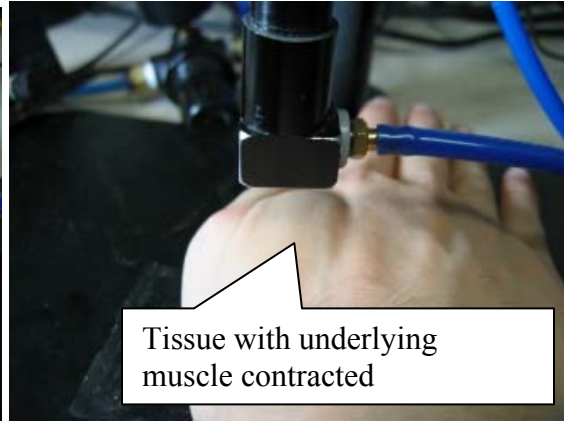
647 Fig. 3



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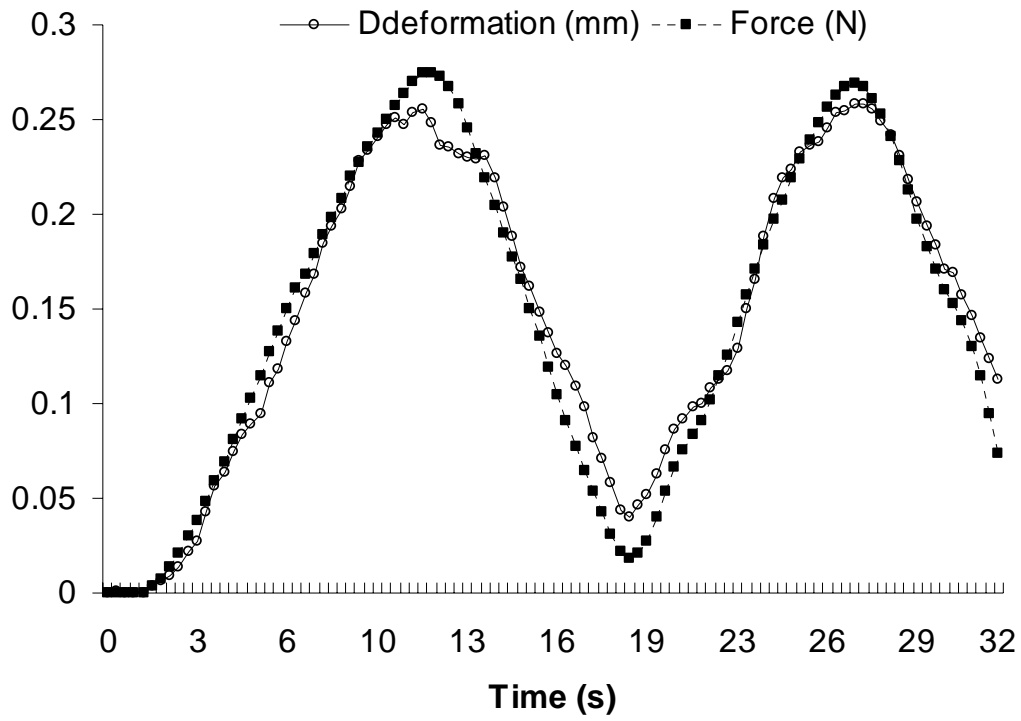
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(a)



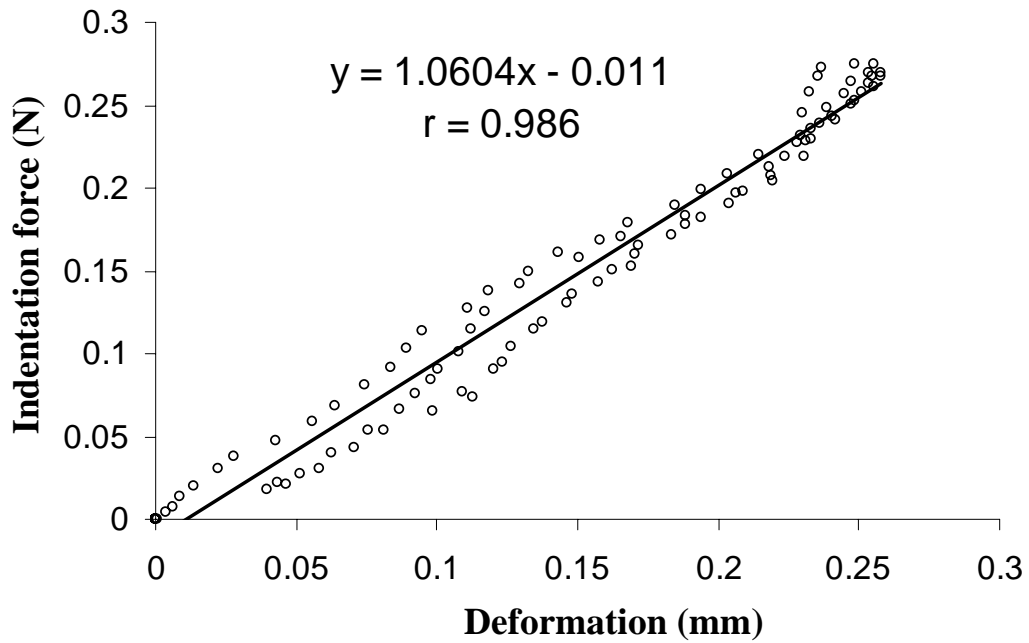
(b)

650 Fig. 4 (a) & (b)



651

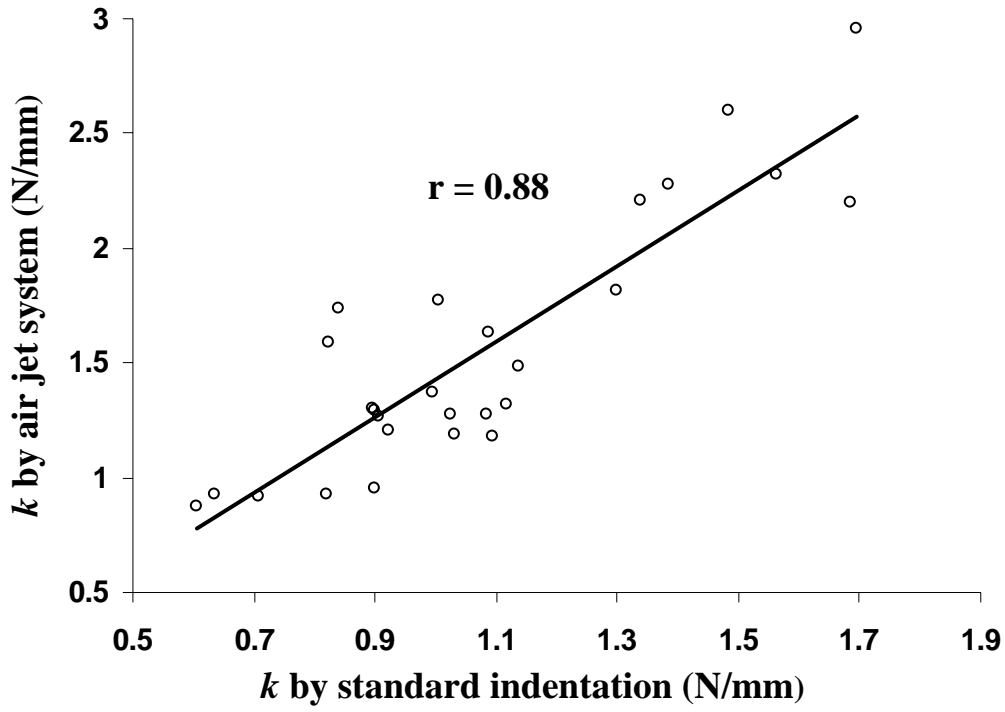
652 Fig. 5 (a)



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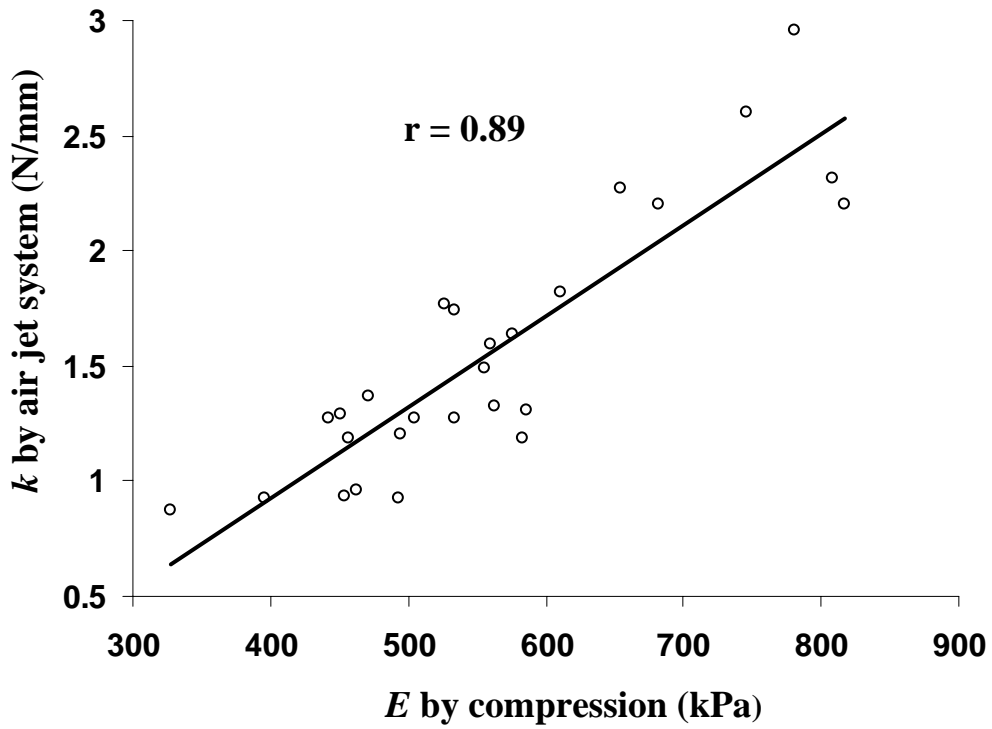
654 Fig. 5 (b)

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656

657 Fig. 6 (a)



658

659 Fig. 6 (b)