



● *Original Contribution*

A NEW METHOD FOR EVALUATION OF FRACTURE HEALING BY ECHO TRACKING

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Abstract—Assessment of bone healing on radiographs depends on the volume and radio-opacity of callus at the healing site, but is not necessarily objective, and there are differences of judgment among observers. To overcome this disadvantage, a clinical system was developed to quantify the stiffness of healing fractures of the tibia in patients by the echo tracking (ET) method in a manner similar to a three-point bending test. The purpose of this study was to ensure that the ET system could clinically assess the progress, delay or arrest of healing. The fibular head and the lateral malleolus were supported. A 7.5-MHz ultrasound probe was placed on the proximal and distal fragments and a load of 25 N was applied. Five tracking points were set along the long axis of the ultrasound probe at intervals of 10 mm. With a multiple ET system, two probes measured the displacement of five tracking points on each of the proximal and distal fragments of the tibia, thereby detecting the bending of the two fragments generated by the load. ET angle was defined as the sum of the inclinations of the proximal and distal fragments. Eight tibial fractures in seven patients treated by a cast or internal fixation were measured over time. In patients with radiographically normal healing, the bending angle decreased exponentially over time. However, in patients with nonunion, the angle remained the same over time. It was demonstrated that the ET method could be clinically applicable to evaluate fracture healing as a versatile, quantitative and noninvasive technique. (E-mail: ohnishi-dis@h.u-tokyo.ac.jp) © 2008 World Federation for Ultrasound in Medicine & Biology.

Key Words: Ultrasound, Echo tracking, Fracture site stiffness, Fracture healing.

INTRODUCTION

The most important issue in assessment of fracture healing is to obtain information about restoration of the mechanical integrity of the bone. In clinical practice, fracture healing is usually judged from serial radiographs. Assessment of bone healing on radiographs depends on the volume and radio-opacity of callus at the healing site, but is not necessarily objective, and there are differences of judgment among observers. In addition, radiographs cannot evaluate fracture site strength. In these respects, assessment of fracture healing by using radiographs is far from ideal.

The stated disadvantages of radiography for assessment of fracture healing have been pointed out in recent years, and various other methods of assessment have been developed. Jernberger (1970) devised an invasive

method for measuring the bending stiffness of healing fractures of the tibia. With his method, the proximal and distal bone fragments were fixed by screws that were connected to a specially designed beam, and a load was applied through a screw at the center of the fixing screws. The method was based on the principle governing the bending of two beams connected at the ends and subjected to a bending force applied at the midpoint. Burny et al. (1984) developed a method that used a strain gauge attached to a fixator shaft. With their method, the strain gauge readings were monitored over time during weight bearing, and the pattern of fracture healing was classified into seven categories (such as normal, delayed, arrested, etc.). Assessment using acoustic emission (AE) was developed by Nicholls and Berg (1981), who detected acoustic pulses generated by microscopic failure of the bone under loading. The investigation by Watanabe et al. (2001) revealed that AE signals occurred with the yielding of callus. However, the strain gauge method and the AE method have the disadvantage that both are limited to patients with external fixation, and both require the in-

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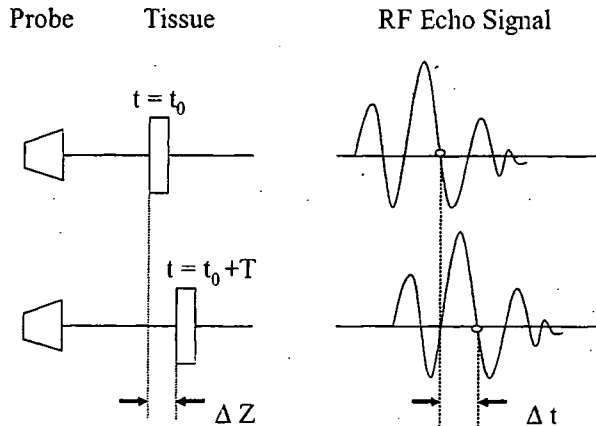


Fig. 1. The target tissue may move closer to or away from an ultrasonic probe over the distance ΔZ during a pulse repetition time of ultrasonic waves (T), causing phase delay of the RF echo signal (Δt). The ET method measures the extent of this displacement by tracking the initialized phase pattern of the echo signal.

section of screw pins or wires. For these reasons, such methods have not been widely used and a new method is needed that is both noninvasive and widely applicable.

To overcome such limitations, we developed a new method for the noninvasive and quantitative assessment of fracture healing. Bone always undergoes deformation in response to an applied load. By quantitatively measuring this deformation, it is possible to assess the mechanical properties of bone and thereby estimate the strength of a fracture site. In this study, we attempted to noninvasively assess the bending stiffness of the healing fracture sites after applying a load. To measure bending stiffness, we focused on ultrasound because it is noninvasive. Precise measurement of the displacement of a specific point can be done by the echo tracking (ET) method. This method is a technique for measuring minute displacement of a certain point on a tissue by detecting a wave pattern in the radiofrequency (RF) echo signal reflected from the target tissue (Fig. 1) (Hokanson et al. 1972). To apply this technique for detection of bone deformation, we improved it so that displacement could be measured with an accuracy of $2.6 \mu\text{m}$ (Matsuyama et al. 2006). We also developed a multi-ET system that was able to simultaneously track dynamic movement at multiple points on the bone surface. In our previous study of the three-point bending test using a porcine tibia, the strain gauge readings and the data from the multi-ET system showed an almost perfect linear correlation with the load ($r = 0.998$). These results indicated the possibility of using the echo tracking method to detect bone surface deformation.

The purpose of this study was to determine whether our newly developed ET system could clinically assess the progress, delay or arrest of healing by detecting the

bending stiffness at the fracture healing site. Fracture healing was evaluated in patients with tibia fracture treated by a cast or internal fixation.

METHODS

A clinical system was developed to quantify the stiffness of healing fractures of the tibia in patients by the ET method in a similar manner to a three-point bending test. Five tracking points were set along the long axis of the ultrasound probe at intervals of 10 mm. With a multiple ET system, two probes measured the displacement of five tracking points on each of the proximal and distal fragments of the tibia, thereby detecting the bending of the two fragments generated by the load. ET angle was defined as the sum of the inclinations of the proximal and distal fragments (Fig. 2). When callus was weak in the initial stage of healing, the tracked points were almost in a straight line and the inclination of the two fragments was calculated directly. However, when the callus was more rigid in the late stage of healing, the line connecting the points was curved and the inclination was obtained from the slope of the linear regression equation for the displacement of the points.

Before clinical application of this method, its accuracy was evaluated by measuring the inclination of the metal flat panel.

Measurement of the accuracy of ET angle using an inclined flat metal panel

A flat stainless steel (SUS 420J) panel (length 270 mm, width 60 mm, thickness 5 mm) was used, which had a parallel accuracy and flatness variation of $<2 \mu\text{m}$. One end of the panel was attached to a magnet stand (DG, Noga Japan Ltd, Saitama, Japan), and the other side was attached to a goniometer (X13-001, Tsukumo Co. Ltd, Saitama, Japan) fixed to another magnet stand. Then, the

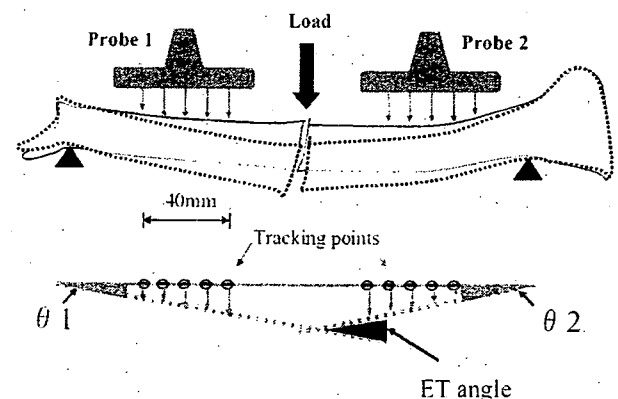


Fig. 2. Probes are set on each of the proximal and distal fragments of the tibia to detect the bending of the two fragments generated by a load. The ET angle is defined as the sum of the inclination of both fragments.

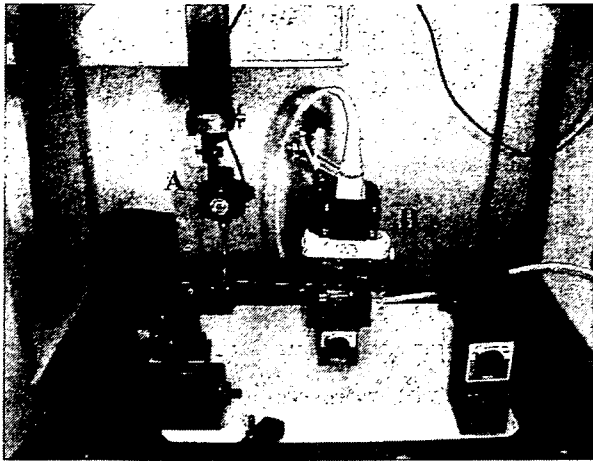


Fig. 3. The accuracy of the ET measurement was evaluated by measuring the inclination of the flat metal panel simultaneously using a 3-D measuring device. (A) 3-D measuring device; (B) 7.5-M Hz linear ultrasound probe.

metal panel was inclined by increasing the height of the goniometer stand. A 7.5-M Hz linear ultrasound probe (UST-5710-7.5, Aloka Co. Ltd., Tokyo) was set at a distance of 20 mm from the panel to measure the changes of displacement of each of five points on the panel (Fig. 3). Using these data, the ET angle of the panel was calculated. At the same time, the inclination of the panel was accurately measured using a 3-D measuring device (AE112, Mitsutoyo, Kanagawa, Japan) with an accuracy of 1 μ m. The panel was inclined by elevating the sliding mechanism of the stand by 0.4 mm and the inclination of the panel was measured 5 times, after which the mean and standard deviation were calculated. Accuracy was evaluated by calculating the standard deviation of the difference between the ET angle and the inclination measured by the 3-D measuring device in each of the measurement trials.

Clinical measurement of fracture site bending stiffness

Eight tibial fractures in seven patients with an average age of 37 y (range 24–69 y) were measured (Table

1). Two fractures of two patients were treated conservatively with a cast, and six fractures of five patients were treated by internal fixation (locked intramedullary nailing in 4, plating in 1 and screws in 1). The average measurement period was 40.8 wk (21–60 wk), and the average number of measurements was 7.5 (5–11).

Patients assumed the supine position with both knees extended, and the affected leg was held horizontal with the antero-medial aspect of the tibia upwards. The fibular head and the lateral malleolus were supported and held tight by a Vacufix (Muranaka Medical Instrument Co., Ltd., Osaka, Japan) to avoid rotation of the leg during loading trials. Before measurement, B-mode images of the short axis of the proximal and distal fragments of the tibia were obtained to identify the center in both directions. By connecting both of the centers, the anatomical axis of the tibia was identified. A 7.5-MHz ultrasound probe was placed on the antero-medial aspect of each of the proximal and distal fragments in the long axis. Each probe was equipped with a multi-ET system with five tracking points at 10-mm intervals. The probes were set vertically on the skin of the leg and held tight with an articulated holder (DG61003, Noga Japan Ltd., Saitama, Japan). A load of 25 N was applied at a rate of 5 N/s and then reduced to 0 N at the same rate using a force gauge (DNP, Imada, Osaka, Japan) parallel to the direction of the probe at the most distal part of the proximal fragment adjacent to the fracture site (Fig. 4). For the initial measurement obtained in each patient, the loading point was set right on the long axis near the fracture site using a B-mode image as a guide. With this setup, the tibia was bent in the same way as for a three-point bending test in the direction of the ultrasound beam. In patients with oblique or spiral fractures, the loading point and the tracking points were set so that they did not cover the fracture site. In patients with a bone graft at the fracture site, the loading point was set on the graft, but the probes were placed so as not to cover it. In the patient with a plate, both the proximal and distal probes were set on the plate surface to measure bending of the plate. Using the multi-ET system, the probes

Table 1. Clinical cases of the tibial fracture

Case	Gender	Age	Limb	Treatment fracture healing	Measurement period (Initial-final)	Radiographic finding
1	F	24	L	Casting	4–47 wk	Normal
2	M	29	R	Casting	7–28 wk	Normal
3	M	23	R	Bone grafting	8–27 mo	Normal
4	M	31	R	Nailing	4–39 wk	Normal
5	F	57	R	Nailing	5–10 mo	Normal
6	F	57	L	Nailing	6–10 mo	Normal
7	F	26	R	Nailing	5 y 2 mo–5 y 7 mo	Nonunion
8	M	69	R	Plating	9–45 wk	Delayed

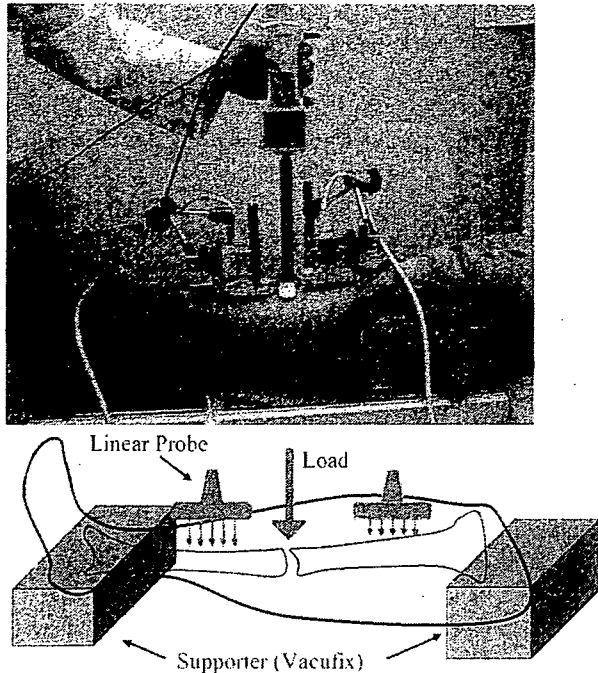


Fig. 4. The affected leg of a patient was held horizontal with the antero-medial aspect of the tibia upwards. The fibular head and the lateral malleolus were supported and held tight by a Vacufix. The probes were set vertically on the skin of the leg and held tight with an articulated arm. A load was applied using a force gauge parallel to the direction of the probe.

detected the angle between the proximal and distal fragments generated by the load. Measurement was repeated five times, and the mean and the standard deviation of the ET angle were calculated.

Fracture healing was assessed at intervals of two or three weeks until radiographic union or arrest of healing occurred. In each patient, the decrease of the ET angle was statistically examined to determine whether it decreased exponentially and whether the decrease was significant. To evaluate the changes of the ET angle over time, exponential regression analysis was performed, and the curve of the ET angle vs. time relation was drawn. Differences were considered significant when the p value was less than 0.05.

To investigate the influence of the position of the probes and the patient on the results, the precision of the method was evaluated by repeated measurement of the ET angle in a patient with a diaphyseal fracture of the tibia treated by a cast (case 2). In addition, the linearity of the relation between the load and the ET angle was assessed by incrementally increasing the load from 10 to 30 N. The ultrasound device (SSD 1000, Aloka Co. Ltd.) used in this investigation is used clinically and its safety has been established. The protocol of this investigation was approved by the ethics committee of The University of Tokyo Hospital, and the patients were enrolled after informed consent was obtained.

RESULTS

Accuracy of ET angle measurement for a flat metal panel

Measurement of the inclination of the flat metal panel showed that the average inclination was 0.117° and the standard deviation was 0.002° . The average inclination obtained with the 3-D measuring device was 0.116° , with a standard deviation of 0.003° . The standard deviation of the differences between the data obtained by the ET method and by the 3-D measuring device was 0.002° .

Clinical measurement of fracture site bending stiffness

The average time required for measurement was 17 min (range 15–20 min). At each loading trial, none of the patients complained of pain and there were no complications related to measurement.

The precision of this method was evaluated by repeating measurement of case 2 (treated with a cast), with repositioning of the leg and the ultrasound probes. The mean and standard deviation of the ET angle were 0.316 ± 0.015 , and the coefficient of variation was calculated to be 4.6%. The linearity of the relation between the load and the bending angle was very high, with a correlation coefficient of 0.997.

Cases presentation

Case 1: A 24-year-old-woman treated with a cast.

The patient sustained a spiral fracture of the proximal diaphysis of the tibia in a traffic accident, and a patella tendon bearing brace cast was applied. Healing was assessed by the ET method, as well as radiographs a total of 11 times from 4 weeks to 47 weeks after fracture. The fracture line became opaque and the callus volume increased from 4 weeks to 19 weeks, but after 26 weeks there was almost no change of the thickness of the callus. On the other hand, measurement showed that the ET angle was about 1° at 4 weeks, and that it decreased exponentially ($y = 1.40e^{-0.105x}$, $r = -0.975$, $p < 0.0001$). The ET angles of both cases 1 and 2 treated with a cast decreased exponentially over time and they reached the level of the intact side by 22 weeks (Fig. 5a, b).

Case 7: A 26-year-old-woman with a fracture of the diaphysis of the tibia treated by a locked intramedullary nailing. ET measurement was performed five times from 5 y 2 mo to 6 y 7 mo after fracture. Her X-ray films showed hypertrophic nonunion, but judgment whether healing was proceeding was extremely difficult. ET measurement showed that there was no significant decrease of the angle over a period of 1 y and 5 mo ($y = 0.264e^{0.002x}$, $r = 0.238$, $p = 0.700$) (Fig. 6a, b).

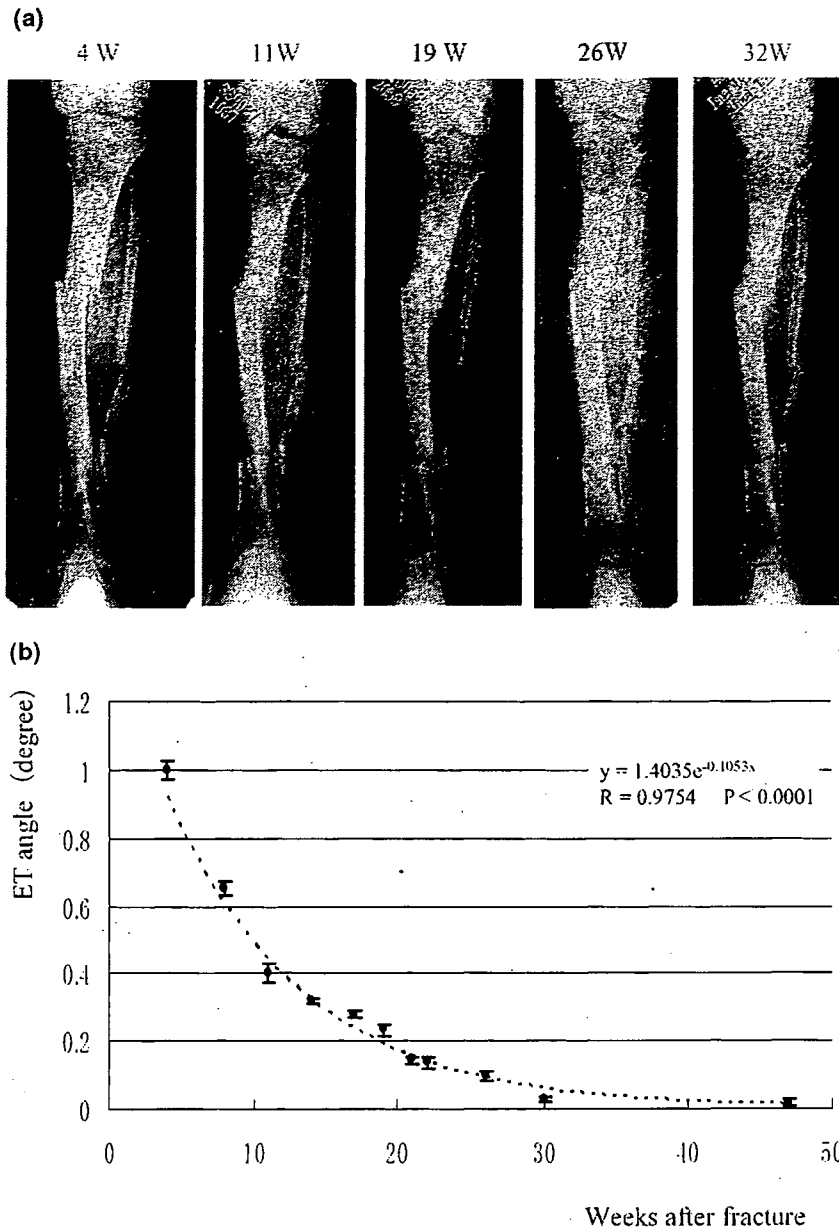


Fig. 5. (a) Time sequential change of the fracture site X-ray from 4 weeks to 32 weeks after fracture in case 1 treated with casting. The fracture site healed normally. (b) In the same patient, the ET angle was plotted. The ET angles decreased exponentially over time.

Case 8: A 69-year-old-man with a long oblique fracture treated with a plate. His X-ray films showed a long oblique fracture line extending for almost 80 mm. Measurement was performed 10 times from 9 weeks to 45 weeks after fracture, during which period almost no change of the fracture site or callus was recognized on X-ray films. The ET method measured the bending angle of the plate. The change was very slow, but the angle decreased significantly from 0.28 to 0.2 degrees, and then finally declined to 0.1 degree. The overall

change showed an exponential curve ($y = 0.40e^{-0.02x}$, $r = -0.895$, $p = 0.0005$) (Fig. 7a, b). In patients with radiographically normal healing, the bending angle decreased exponentially over time (Fig. 8). However, in patients with nonunion, the angle remained the same over time.

DISCUSSION

Our method allows noninvasive assessment of bending stiffness at the healing site, so it can be appli-

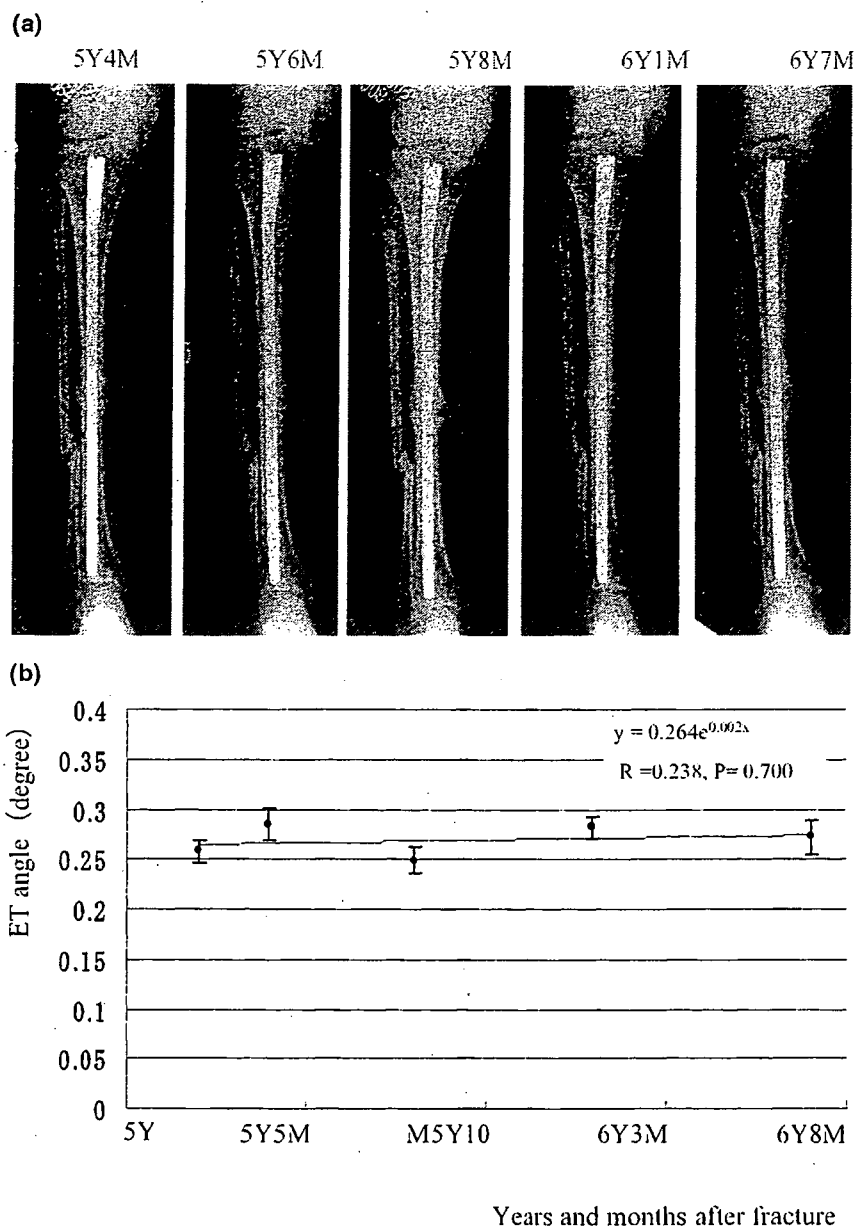


Fig. 6. (a) Time sequential change of the fracture site X-ray from 5 y 4 mo to 6 y 7 mo after fracture in case 7, treated with intramedullary nailing. The X-ray films showed hypertrophic nonunion, but judgment of whether healing was proceeding was extremely difficult. (b) In case 7, the ET angle showed no change over time and the regression lines showed no significant decrease.

cable to patients treated conservatively as well as those managed by surgical intervention with plating or intramedullary nailing.

In this study, the precision and reproducibility of the method were evaluated. The precision of measuring displacement by using the echo tracking system specially designed for bone surface measurement has already been assessed, and a precision of 2.6μ was demonstrated in our previous study. However, the precision of measuring the bending angle has not been investigated before. We

obtained a precision of 0.002° , which was thought to be adequate based on the results of the study by Moorcroft et al. (2001) that evaluated fracture healing. They used the three-point bending test to generate angles of 0.4 to 1.0° in an *in-vivo* measurement trial and connected a goniometer to the bone fragment *via* screw pins fixed to a side bar of the external fixator to detect bending at the fracture site.

When estimation of the linearity of measurement was done in relation to the load, there was excellent

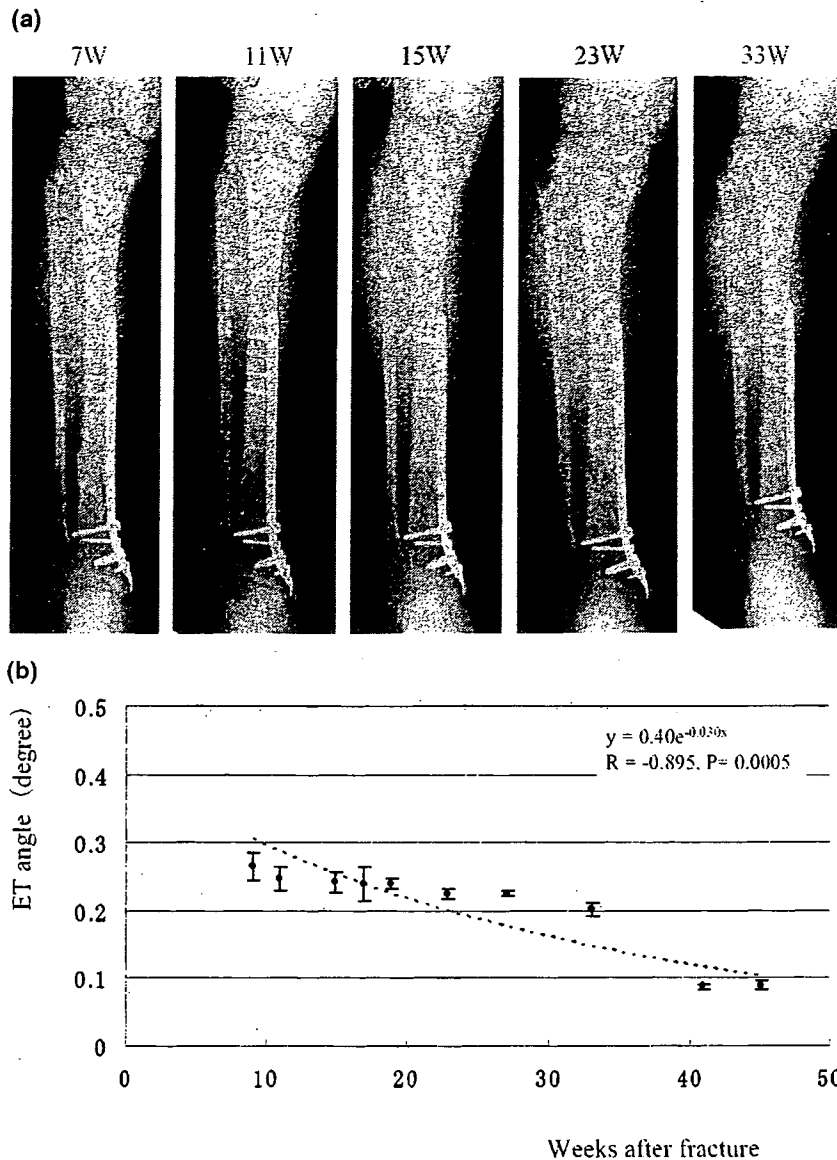


Fig. 7. (a) The X-ray films of case 8, treated with plating. No change of the fracture site or callus was recognized on X-ray films. (b) The ET method measured the bending angle of the plate. The change was very slow, but the angle decreased significantly from 0.28 to 0.2°, and then finally declined to 0.1°.

linearity between magnitude of the load and the ET angle ($r = 0.997$), indicating that elastic deformation of the fracture site had occurred under a load range of 10 to 30 N. Therefore, measurement was shown to be noninvasive as well as safe, without causing any residual deformity.

Reproducibility of the measurement method was estimated to be 0.015° , which was adequate to evaluate fracture healing quantitatively, because the angle ranged from around 1° in the initial stage to about 0.1° in the final stage when it was almost equivalent to that of the intact tibia. However, we have to improve the reproducibility of measurement *in vivo*. The factors affecting reproducibility *in*

in vivo include the position of the leg, loading direction and positions of the probes. Among these, the positioning or fixation of the leg seems to have the most influence on the reproducibility of measurement.

For clinical evaluation of fracture healing, data obtained by the ET method were compared with X-ray findings over time. In patients with delayed healing or nonunion, judgment of the healing process using X-ray films was difficult because the direction and conditions of obtaining images were not exactly the same every time, so the findings were not reproducible. In contrast, the echo tracking method evaluated fracture stiffness

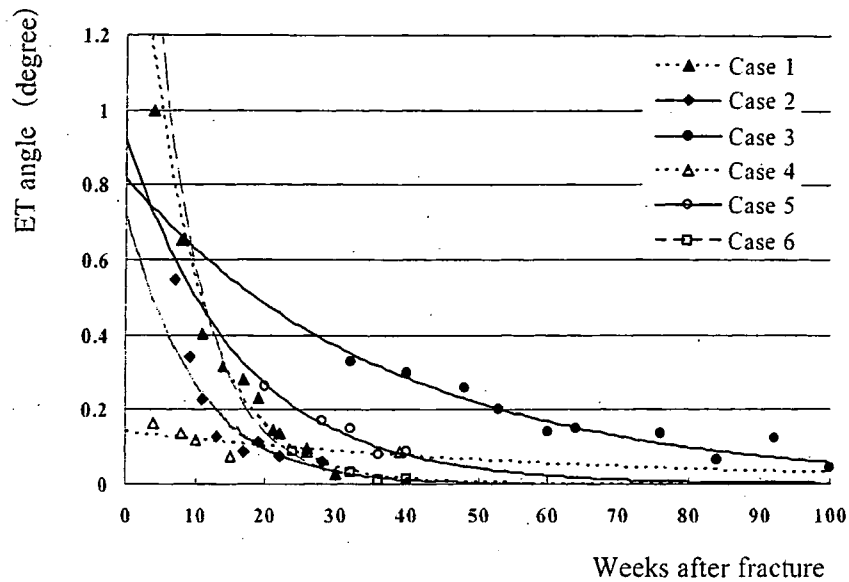


Fig. 8. In cases 1 through 6, the changes of the ET angle showed an exponential pattern. The correlation coefficients obtained by the regression equation for the ET angle and time were very high in these cases.

with considerable accuracy, sensitivity and reproducibility.

In patients with radiographically normal healing, the bending angle decreased exponentially over time. However, in patients with nonunion, the angle remained the same over time. According to the results obtained with previous methods such as the strain gauge method and the invasive method of Jernberger (1970), strain or deformation caused by loading at the healing site has been reported to diminish exponentially over time in patients with normal healing. Among these previous studies, Bourgois and Burny (1972) evaluated fracture healing in hundreds of patients treated with an external fixator that was instrumented with a strain gauge. They not only accumulated considerable clinical data on the strain readings over time, but also theoretically proved by mathematical simulation that the change of the strain over time during normal healing could be expressed as a typical hyperbolic curve. In addition to this, they proved that the time course of the change in strain could also be a hyperbolic curve by developing fracture simulation models with stabilization by intramedullary nailing, plating and external fixation. As a result, their clinical data were compatible with those for the theoretical model of external fixation. They classified the pattern of fracture healing into seven categories depending on the difference in the healing process. Among them, normal healing was defined as healing in which the strain reading vs. time curve reaches a plateau at 60 to 90 d after fracture. Slow healing was defined as healing in which the decline of strain was very slow compared with the normal pat-

tern but the healing process was progressive over time. Nonunion was defined as cessation of the progress of healing. In two patients treated with a cast in our study, the ET angle decreased rapidly until 10 weeks after fracture to a level twice that on the intact side, and then it decreased slowly. The exponential regression curve for the echo tracking angle vs. time showed a very strong correlation (case 1, $r = -0.975$). Therefore, it can be concluded that the echo tracking method could be used to evaluate normal healing as proposed by Burny et al. (1984). As shown in Fig. 5, the progress of healing in patients treated with intramedullary nailing and bone grafting could be assessed by using the ET method. The ET angle vs. time relation in these cases was also expressed by exponential curves. However, the ET angle curve of patient 7 (Fig. 6b) did not show any significant decrease of the angle and there was no correlation between the ET angle and time. From this, the healing process was diagnosed as nonunion. The ET angle of patient 8, treated with plating, showed an extremely slow decrease over time from 9 weeks to 33 weeks, but reduction of the angle was statistically significant until 45 weeks, so the healing process was concluded to be delayed.

Fracture site stiffness was adopted as a parameter for evaluation that was thought to be correlated with strength of bone healing. In various earlier studies of fracture site mechanical properties, stiffness was measured to estimate the strength of the fracture site. However, stiffness is not necessarily correlated with strength. Chegade et al. (1997) investigated this relationship in 24

sheep. The tibia was stabilized with an external fixator and then osteotomy was done. Next, the tibiae were excised at 6, 8 and 10 wk after osteotomy and a 4-point bending test was done. As a result, in the initial stage of healing, stiffness showed a strong correlation with strength ($r = 0.89$), but there was no correlation between them in the remodeling stage. However, as Chehade *et al.* (1997) stated, because the stiffness of the fracture site is strongly correlated with the strength until remodeling is initiated, it is clinically significant to monitor fracture site stiffness as a substitute for strength to determine the appropriate level of weight bearing so that patients can avoid refracture because of overloading the fracture site during postoperative management. In the remodeling stage, we need to pay special attention to the relationship between stiffness and strength, even if stiffness reached the same value as the intact side.

Fracture healing was evaluated quantitatively by the echo tracking method in patients treated conservatively as well as by internal fixation. All previous methods of assessment could only be applied to patients treated with an external fixator that required the insertion of wires or screw pins, and none of the methods could achieve evaluation in a totally noninvasive manner. The potential problem with evaluating patients treated with internal osteosynthetic devices such as intramedullary nails or plates is that the stiffness at the fracture site is the sum of stiffness for both the healing fracture and the implant. The stiffness of the implant is very high compared with that of the healing fracture because it is made of a metal such as stainless steel or titanium-aluminum-vanadium alloy. Therefore, the combined stiffness at the fracture site is usually very high compared with that in patients receiving conservative treatment by casting. In such patients with internal osteosynthetic devices, comparison of stiffness with the intact side does not have any meaning

for evaluation of fracture healing. Therefore, we have to be careful with interpretation of the changes of stiffness over time in such cases. How the implanted material and the configuration of stabilization affect fracture site stiffness should be investigated in the future so that we can assess fracture healing more precisely in patients with internal fixation.

In conclusion, it was demonstrated that the echo tracking method could be clinically applicable to evaluate fracture healing as a versatile, quantitative and non-invasive technique. Further development of this method should be performed so that it can be applied to other anatomical sites by improving accuracy and precision.

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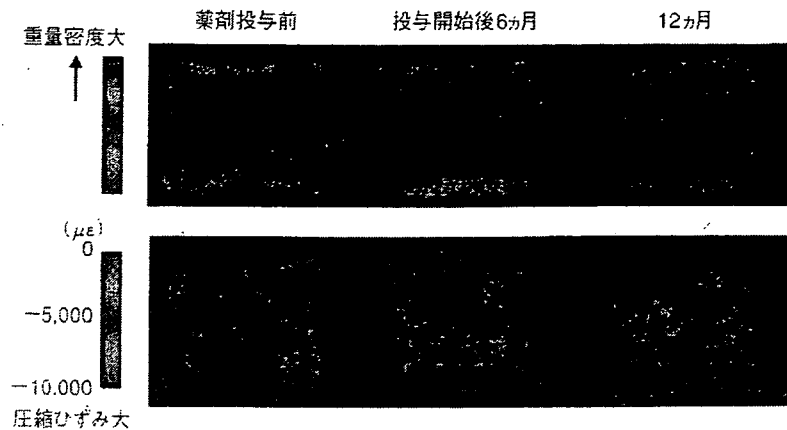


図2 74歳女性, アレンドロネート単独投与に伴う第2腰椎の重量密度および最小主ひずみ分布の変化

度解析値の解離がみられていた。

CT/有限要素法は骨強度だけではなく、骨力学特性を評価することが可能である。アレンドロネート投与開始6ヵ月でDXAによる骨密度が-0.9%低下に対しCT/有限要素法による椎体強度が33.3%増加、12ヵ月で骨密度が-0.3%低下に対し椎体強度が59.3%増加と、骨密度と椎体強度が解離した症例では、皮質シェルと隣接する海綿骨内の重量密度増加がみられ、最小主ひずみ分布で薬剤投与前では圧縮ひずみが著しく骨折が生じやすい領域に薬剤投与後では圧縮ひずみの減少がみられていた(図2)。

薬剤効果として、皮質シェル近傍の重量密度が増加するような骨基質分布の変化が起こり、圧縮応力の集中が軽減して最小主ひずみ分布が改善し、椎体強度が増加したと考えられ、DXAによる骨密度でとらえられない薬剤効果をCT/有限要素法で評価できる可能性が示唆された。

CT/有限要素法は骨密度に加え、ジオメトリ(geometry)、骨基質分布を含めて評価することが可能である。一方、CTの空間分解能に依存して骨微細構造や皮質骨の詳細な評価、マイクロダメージ(microdamage)の評価には限界があり、また骨代謝を評価することができない。さらに、DXAやpQCTに比較して放射線被曝量が多い点が検討すべき課題である。

おわりに

骨粗鬆症患者に対するアレンドロネート投与で、CT/有限要素法による椎体強度は、DXAによる骨密度より早期の投与開始後3ヵ月で薬剤効果を反映していた。薬剤効果として、骨密度増加だけではなく、骨基質分布の改善、骨折が生じやすい領域の最小主ひずみ分布の改善がみられていた。CT/有限要素法は、骨強度・骨力学特性の評価が可能という点で、骨粗鬆症治療薬の効果判定に有用と考えられる。

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直達式骨折整復の臨床データ取得システム

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A Clinical Data logging system of Direct fracture reduction.

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Abstract: Clinical data quantification of a fracture reduction has become important as the development of new fracture reduction technology such as navigation and robot assisted fracture reduction. A reduction force and a reduction path are the key points in considering the control and safety of these new reduction methods. We have developed a clinical data logging system(CDLS) of direct fracture reduction, which reduces using ring-frame connected bone fragment directly. The CDLS synchronously records the reduction force, the reduction path and two video signals. One records the images from C-arm and the other records whole surgery process. This paper introduces the structure of CDLS and the resultants of application to fracture model.

Key words: Clinical data, Fracture reduction, Reduction force, Reduction path.

1. 序論

骨折治療は近年大きく発展してきており、低侵襲で整復の精度を上げることが可能となってきた。特に最近、術中に骨片の位置が認知できるナビゲーションシステムを用いた治療方法に関する研究が盛んに行われている[1]。骨折治療法は様々であるが、どの治療法でも整復を伴い整復の結果が治療の成果に大きく影響する。大腿骨頸部骨折の整復に関しては大きい整復力が必要で術者への負担が過大になる。そこで、ロボットを用いた整復支援システムに関する研究が報告されている[2, 3]。このような整復支援ロボットには過渡の整復力を防ぐための安全装置が装着されている。一定以上の整復力が働いたときにその軸がフリー状態になるフェイルセーフ装置が一例である[3]。

ロボットの制御方法と安全装置の仕様を決めるためには、整復手術中の整復パス、整復力などの臨床データを参照するべきであり、今までは大きく必要とされていなかった骨折整復術中の臨床データを記録し定量化する必要性が整復技術の発展と共に高くなってきた。このため、Thomasらは整復力を定量化した[4]。しかし、整復パスを記録していないため骨片の位置と整復力との関係は解析されていない。そこで我々は骨片に打ったピンをリングフレームで固定し、そのフレームを持ち骨折整復を行う直達式整復術中の整復力とパスを記録するシステムを開発した[5]。しかし、整復力を計測するセンサの滅菌を検討しなかつたので、臨床使用には適してなかつた。本研究では、臨床使用可能とするように改良し、さらに画像入力

も可能としたシステムを紹介し、模擬骨を用いた整復への応用結果を報告する。

2. 臨床データ取得システム

臨床データ取得システム(Clinical Data loggings system, CDLS)では骨折整復時に発生する整復力、整復パス、C-arm画像と現場の様子を記録することが望ましい。我々の開発した CDLS は、PC、画像入力ボード(PCI-5531, Interface)、整復力を測るための力覚センサ(IFS-67M25T50, Nitta, 定格 Fx, Fy [N]: 450; Fz [N]: 900; Mx - Mz [Nm]: 40)、整復パスを追跡するための光学式 3 次元位置計測装置(Polaris, NDI, Canada)で構成され、以上のデータを計測可能とした。これらのデータは位置と整復力との関係などを分析するため、同期を取って記録する必要があり、システムではひとつのプロセスでコンピュータ側からデータ要求信号を送りそのときのデータを取得する方法を用いた。

Thomasらの研究結果によると骨片の牽引力の最大値は396N、回旋トルクの最大値は74Nmなので使用した力覚センサの定格は十分である。しかし、力センサの適切な滅菌法がないため、滅菌可能なカバーを製作した。カバーはアシスタントの補助により、術者の手を汚すことなく、取り付けられる。Fig. 1にカバーを取り付けた力覚センサと整復装置の外観を示す。図の右はカバーを外したときのセンサとカバーの様子である。カバーには整復パスを記録するため、光学式 3 次元位置計測装置のマーカを付着した。骨片の位置と整復力を計算するための概念図を

Fig. 2 に示す. 光学式 3 次元位置計測装置, 力覚センサ, C-armと骨片の座標系を各々 ΣP , ΣS , ΣC , ΣB と設定する. ΣP から ΣS と ΣC への変換行列 ${}^P T_S$ と ${}^P T_C$ は 3 次元計測器から計測できる. ΣC から ΣB への変換行列 ${}^C T_B$ は 2D/3D レジストレーションなどの手法で求められる. 以上の変換行列で力覚センサの座標系から骨座標系への変換行列 ${}^S T_B$ が式(1)で計算できる.

$${}^S T_B = ({}^P T_S)^{-1} {}^P T_C {}^C T_B \quad (1)$$

そこで骨片と整復装置を剛体として考えると, ${}^S T_B$ を一回計算しておくことにより, 整復中はセンサのマーカを追跡することだけで光学式 3 次元位置計測装置から見た骨の位置と姿勢が式(2)で求められる.

$${}^P T_B = {}^P T_S {}^S T_B \quad (2)$$

整復力に関しても, 式(3)(4)により, センサから計測した力が骨座標系の力に変換できる.

$${}^B T_S = ({}^S T_B)^{-1} = \begin{bmatrix} {}^B R_S & {}^B P_S \\ 0 & 1 \end{bmatrix} \quad (3)$$

$$\begin{bmatrix} \text{bone } \mathbf{f} \\ \text{bone } \mathbf{n} \end{bmatrix} = \begin{bmatrix} {}^B R_S & 0 \\ {}^B P_S \times & {}^B R_S \end{bmatrix} \begin{bmatrix} \text{sensor } \mathbf{f} \\ \text{sensor } \mathbf{n} \end{bmatrix} \quad (4)$$

3. 骨折モデルの整復への応用

骨折モデルは X 線で確認できる材質のモデル骨を骨折させ, 筋肉代わりにゴムを貼り付けることで製作した. 大腿骨頸部骨折の場合, 骨片は中殿筋, 外閉鎖筋などの働きにより上半身のほうに引っ張られ, 外旋されるのでその働きをゴムで模擬した. 今回の実験では骨の座標はペンマーカを用いて指定した. 整復は整形外科医により行い, そのときのデータを CDLS により取得した. C-armは使わず, 代わりにカメラからの画像を入力した.

データの取得周期 0.3s で 2 枚の画像, 整復力と整復パスを記録することができた. Fig. 3 に取得した整復力を示す. 整復力は y 軸の力(牽引方向)とモーメント(内旋方向)に大きく必要とすることが解る. 整復パスはマーカが術者に隠され一部記録できてない部分があった.

4. 考察と結論

実験では模擬骨を用いたので, 計測した整復力の大きさには意味がないが, どの成分の整復力が働くか確認することができた. また整復時の骨片の動きも確認することができた. センサのマーカは術者に邪魔にならない横方向に移すべきである. データ取得周期は 0.3 秒で遅いが, 整復の動作がゆっくり行われるので問題にはならない. 周期をあげるためには複数のパソコンにより並列処理をすればよいが, システムが大きくなるので手術現場では適さない.

今後は, C-armとレジストレーション方法を用いた模擬整復でシステムを評価し, 実際に臨床でデータ取得を行う予定である.

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Fig. 1 Fracture reduction device.

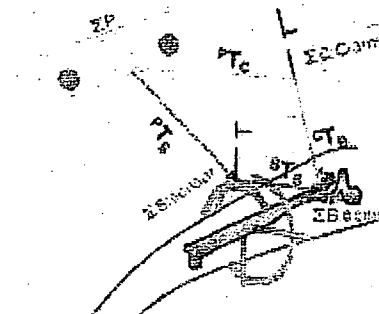


Fig. 2 Coordinate and translation matrix.

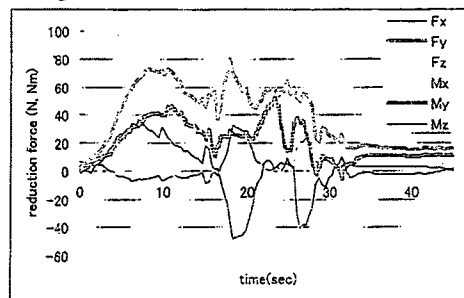


Fig.3 The reduction force during fracture reduction

直達式骨折整復を支援する骨折整復システムの開発

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骨粗鬆症がある患者におこりやすい骨折のなかでも大腿骨頸部骨折は寝たきりとなる可能性の高い骨折である。社会の高齢化の進行に伴い骨粗鬆症の患者が増加すると、大腿骨頸部骨折の患者も増加すると予測されている。

大腿骨頸部骨折の治療法は外科的な手術によるものがほとんどである。手術では大腿骨の遠位骨片を牽引しながら位置決めし、ピンによって固定する。しかし、大腿筋などの周辺組織が萎縮した状態では整復のための牽引に大きな力が必要となるため術者にとって負担となる。また、X線透視下で2次元の情報を用いて位置決めを行わなければならないため、術者の熟練が必要であり、手術を行うことで術者が受けるX線被曝も問題である。

これらの問題に対し、我々は骨折整復システムの開発を行った(Fig. 1)。システムは骨折整復ロボットとナビゲーションシステムに構成されている。骨折整復には、足首をつかみ遠位骨片の位置を合わせる介達式骨折整復方法と、骨片にピンを打ち、ピンに連結されたリングを持って直接整復を行う直達式骨折整復方法がある。骨折整復システムは介達式骨折整復に対して使えるように構成されているが、ロボットを使う利点を考えると骨片の正確な位置決めが可能な直達式骨折整復にも使用されるように構成する必要があり、今回は新たに直達式骨折整復を直達式骨折整復ロボットに実装を行った。

骨折整復ロボットは並進3自由度と回転3自由度の6自由度を有する。骨片の牽引と回旋をするときの整復力が設定値より大きくなると各関連軸をフリーにするフェイルセーフ装置が装着されており安全性を保つ。動作モードはタッチパネルを用いたジョグモード、術者の整復力をパワーアシストする自動モード、ナビゲーションからの指令により自動で整復を行う自動整復モードがある。

直達式整復では、骨とロボットの手先は専用のジグで繋がっているため、骨の長軸とロボットの牽引軸が一致しない。骨片の姿勢だけを変えるため、骨折断面の中心を仮想中心と見なし、ロボットを制御する拘束パワーアシストを実装し、有効性を検証した。

ナビゲーションシステムは術前にCTからの3次元モデルを用いて整復ゴールを計算する。術中にはC-armで撮った画像と3次元モデルをレジストレーションすることにより、実空間での骨片間の位置関係を認知する。骨片の現在位置からゴールまでの整復パスは術者の意見を反映して作成され、整復ロボットに指令を送り整復を行う。構成したシステムは骨折モデルでの整復実験でその有効性を示す。



Fig.1. 骨折整復システム

A Parallel Implementation of 2-D/3-D Image Registration for Computer-Assisted Surgery

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Abstract

This paper presents the design and implementation of a parallel two-dimensional/three-dimensional (2-D/3-D) image registration method for computer-assisted surgery. Our method exploits data and speculative parallelism, aiming at making computation time short enough to carry out registration tasks during surgery. Our experiments show that exploiting both parallelisms reduces computation time on a cluster of 64 PCs from a few tens of minutes to less than a few tens of seconds.

1. Introduction

Image registration is a technique for finding point correspondences between two different images taken usually at different times, from different viewing points, and/or in different modalities. This technique plays an increasingly important role in surgery. For example, registration of preoperative images to intraoperative images is essential to perform image-guided and robot-assisted surgery [6], which minimizes surgical complications for better surgical outcomes.

For this purpose, many researchers have tackled the problem of 2-D/3-D registration [1], which estimates the location and orientation of a 3-D volume with respect to the patient coordinate system using one or more 2-D projected images. The reason why aligning a 3-D volume to 2-D images is due to the limitations of current 3-D imaging systems, such as computed tomography (CT) scans, which have more spatial information but require more acquisition time and radiation exposure, as compared with 2-D imaging systems. Due to these limitations, the intraoperative data is usually 2-D X-ray fluoroscopy or ultrasound images, whereas the preoperative data is 3-D CT volumes, motivating us to deal with 2-D/3-D registration.

One key challenge for 2-D/3-D registration is to develop a fast, accurate, and robust algorithm. Prior algorithms can be classified into two groups: feature- and intensity-based approaches. The intensity-based approach [1, 5, 6] has been shown to provide more accurate and robust results than the

feature-based approach [3, 7]. The intensity-based approach compares the intensity values between the real projected image and the digitally reconstructed radiograph (DRR) generated from the volume. It requires a large amount of computation to iteratively produce DRRs until a best match between the projected image and the DRR is found. Therefore, acceleration strategies are required to minimize surgical time.

In this paper, we present the design and implementation of a parallel 2-D/3-D registration method, aiming at achieving fast, accurate, and robust alignments for computer-assisted surgery. Our method parallelizes an intensity-based algorithm to reduce computation time without degrading the quality of alignment. The key contribution of our method is to provide fast and robust alignments by means of data-parallel and speculative processing, respectively.

2. Intensity-Based 2-D/3-D Registration

The intensity-based algorithm employed in our method has the following three advantages: (1) automated registration by comparing a real projected image and a DRR [1]; (2) robust registration using an information-based similarity measure [5]; and (3) accurate registration using biplane 2-D images [2] and region of interest (ROI) [6]. Due to the limited space, we describe only advantages (1) and (2).

Before describing each advantage, we first define the 2-D/3-D registration problem. To make it easier, we present a definition for a single image rather than biplane images. Given a volume V and a real projected image I_F (see Figure 1), the 2-D/3-D registration technique computes the rigid transformation parameter \mathbf{T} that relates the coordinate system of the volume V and that of the imaging (patient) environment. Here, the rigid transformation is given by $\mathbf{T} = (T_X, T_Y, T_Z, \theta_X, \theta_Y, \theta_Z)$, where the first and last three parameters are the translations and rotations of V .

Figure 2 briefly presents the intensity-based algorithm. The algorithm resolves the registration problem into an optimization problem. That is, in order to register the volume V to the 2-D image I_F , the algorithm optimizes a cost function C associated with the location and orientation of V ,

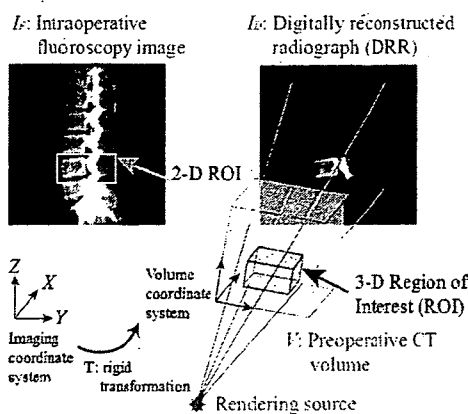


Figure 1. Overview of intensity-based 2-D/3-D registration. In this case, it aligns a CT volume of a real spine to a fluoroscopy image of the spine.

where C represents the similarity measure between the 2-D image I_F and the DRR I_D , generated from V . Furthermore, this optimization is performed in a hierarchical manner in order to reduce the amount of computation. This hierarchy is controlled by the step size λ of the optimization.

The algorithm consists of the following technologies.

DRR generation: As illustrated in Figure 1, a ray casting method generates the DRR I_D . Image intensity $I_D(i, j)$ at point (i, j) on the DRR I_D is computed by accumulating the intensities of the voxels that ray $r(i, j)$ penetrates, where $r(i, j)$ represents a ray that penetrates point (i, j) from the rendering source.

Similarity measure: We use gradient correlation (GC) for our algorithm according to an experimental study [5] on six similarity measures. Although the study found that pattern intensity [6] and gradient difference were the most robust measure for their registration scenario, these measures require intensity correction [5] to minimize the difference between the two images I_F and I_D , because they use a difference image created by subtracting I_D from I_F . In contrast, GC focuses on edge information in the images, so it essentially does not require intensity correction to minimize the difference. Furthermore, the study also showed that GC provided a small failure rate (5%) for clinical datasets, and it was the most robust measure that assumed no intensity correction. Therefore, we use GC as the similarity measure between the two images I_F and I_D : $C(\mathbf{T}) = G(I_F, I_D)$. Here, GC $G(A, B)$ between images A and B is given by $G(A, B) = N/2 (\partial A/\partial i, \partial B/\partial i) + N/2 (\partial A/\partial j, \partial B/\partial j)$, where $\partial A/\partial i$ and $\partial A/\partial j$ ($\partial B/\partial i$ and $\partial B/\partial j$) are the gradient images of A (B , respectively), representing the derivative of the intensity in the horizontal and vertical axes of the image, and $N(A, B)$ is normalized cross correlation (NCC) defined over two images A and B .

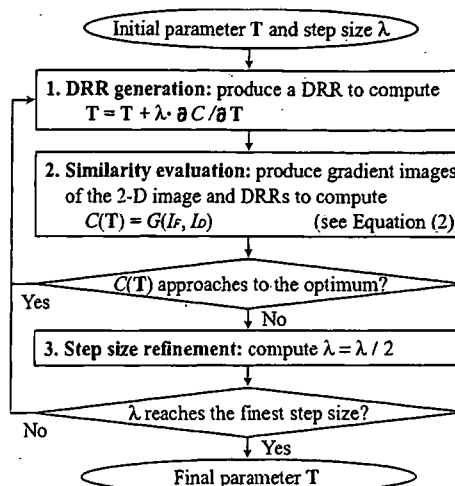


Figure 2. Intensity-based 2-D/3-D registration algorithm.

The gradient images are produced by means of the first derivative of a Gaussian. This filter has the advantage that it reduces and smoothes noise in the images, improving the robustness of alignment. In summary, the intensity values at point (i, j) on the gradient images $\partial A/\partial i$ and $\partial A/\partial j$ are given by convolution with the first derivative Gaussian filters $F_I(i, j)$ and $F_J(i, j)$: $\partial A(i, j)/\partial i = F_I(i, j) * A(i, j)$ and $\partial A(i, j)/\partial j = F_J(i, j) * A(i, j)$, where $F_I(i, j) = \frac{-i}{2\pi\sigma^4} e^{-(i^2+j^2)/2\sigma^2}$, $F_J(i, j) = \frac{-j}{2\pi\sigma^4} e^{-(i^2+j^2)/2\sigma^2}$, and σ is the standard deviation of the distribution and is proportional to the kernel size, namely the size of neighborhood on which the filter operates.

Summarizing the above description, the main computation phases of the algorithm are (a) DRR generation, (b) gradient image generation, and (c) NCC computation.

Optimization: In order to find the optimal transformation parameter \mathbf{T} that maximizes the cost function C , the algorithm employs the steepest descent optimization technique during registration process: $\mathbf{T} = \mathbf{T} + \lambda \partial C/\partial \mathbf{T}$. This optimization stops if a local optimum has been found. The gradient $\partial C/\partial \mathbf{T}$ of the cost function is estimated by using the finite-difference approximation. Because the transformation \mathbf{T} consists of six independent parameters, the computation phases (a)–(c) are repeated 13 times to approximate the gradient $\partial C/\partial \mathbf{T}$ at each optimization step: one repetition for current transformation \mathbf{T} and 12 repetitions for finite differences $\mathbf{T} \pm \lambda \Delta$ of each parameter.

3. Parallelizing 2-D/3-D Registration

In this section we present the design and implementation of our parallel method.

3.1. Design Aspects

To accelerate the registration process, we can exploit three parallelisms as follows.

- **Speculative parallelism:** In the registration algorithm, speculative parallelism can be exploited by simultaneously processing the same registration task with different initial parameters. This is important to prevent unsuccessful registrations (due to local optimums), because the surgery cannot progress until the alignment has correctly finished. Otherwise, the surgery must be performed without the surgical plan. To prevent such undesirable situations, an appropriate transformation is required as the initial parameter T . However, in general, initial parameters are experimentally determined according to the surgeon's experience. Therefore, speculative processing contributes to improve the robustness and confidence of our method.
- **Data parallelism:** Exploiting this parallelism accelerates a single registration task. It can easily be established by using image parallelism [4], where processors take the responsibility for each subtask associated with a small part of the 2-D image. The details are presented later in Section 3.2.
- **Task parallelism:** This parallelism also contributes to accelerate a single registration task. It exists in the finite-difference approximation, where the computation phases (a)–(c) are repeated 13 times. However, this means that the speedup derived by this parallelism is limited by a small factor of 13. Furthermore, load balancing is probably not easy if it is exploited, because 13 cannot divide the number of processors, usually chosen to be a power of two.

From the above discussion, we have decided to exploit speculative and data parallelism.

In addition to the computation phases (a)–(c), input/output (I/O) operations also might become a performance bottleneck after parallelization. However, I/O issues are not critical in our cluster environment for the following two reasons. Firstly, the largest input data, namely the volume V , is the preoperative data, which can be distributed to processors before surgery. Therefore, we can assume that processors have loaded it into their local memory when registration tasks are submitted. Secondly, the remaining data I_F is small enough to be broadcasted rapidly in our cluster. Therefore, we assume that all processors have the entire data, V and I_F , in their local memory.

3.2. Workload Distribution

We now present how our method exploits image parallelism. A good solution to this issue balances workload among processors and minimizes the amount of messages transmitted between processors and the number of sends and receives. To find such a solution, we first investigate the

characteristics of computation phases (a)–(c) with respect to available parallelism, load balancing, and data access pattern. Table 1 shows these characteristics with a preliminary timing result measured on a single processor machine.

- **DRR generation:** The intensity value at any point (i, j) can independently be computed with the values at other points, because different rays can cast independently. The workload associated with each point is nonuniform due to the different number of penetrated voxels. Points around the DRR edge tend to have less workload. In addition to this image parallelism, we also can use object parallelism [4], where processors take the responsibility for each subtask associated with a small part of the volume and then merge locally rendered DRRs into a final DRR. This object-parallel scheme allows processors to load only a small portion of the volume, but it requires communication to generate the final DRR. As mentioned earlier, we assume that all processors have the entire volume, so that our method uses an image-parallel scheme to prevent communication in this most time-consuming phase.
- **Gradient image generation:** As same as in DRR generation, different points can independently be processed to obtain their intensities on the gradient image. The convolution for point (i, j) requires all intensities $A(i + \alpha, j + \beta)$ such that $-[K/2] \leq \alpha, \beta \leq [K/2]$, where K denotes the kernel size of the filter. Note here that this means any point on the gradient image requires DRR generation of its surrounding $K \times K$ neighborhood, because the gradient images are generated from the DRR I_D as well as the image I_F . With regard to load balancing, this computation phase has uniform workload, because the same kernel size K is used for every point. Note also here that the kernel size K is usually a relatively large number, which increases the amount of messages under an inappropriate workload distribution scheme. For example, we use $K = 19$ pixels for typical 2-D ROI sizes ranging from 200×200 to 400×400 pixels.
- **NCC computation:** NCC computation can be approached as a reduction problem, because it can be rewritten as $N(A, B) = \frac{\sum_{i,j} (A(i, j)B(i, j) - \bar{A}\bar{B})}{\sqrt{\sum_{i,j} (A(i, j)^2 - \bar{A}^2)} \sqrt{\sum_{i,j} (B(i, j)^2 - \bar{B}^2)}}$. This equation indicates that NCC can be computed from six local sums: the local sums of the number of points; intensities $A(i, j)$; $B(i, j)$; squared intensities $A(i, j)^2$; $B(i, j)^2$; and multiplied intensities $A(i, j)B(i, j)$. These sums can independently be computed if processors are responsible for the same point (i, j) on images A and B . The workload is perfectly balanced if the same number of points is assigned to each processor. However, communication is required to reduce local sums into a global sum.

According to the analysis mentioned above, our method employs a 2-D block distribution scheme with the overlap

Table 1. Summary of computation phases.

Computation phase	Parallelism	Workload	Data required for intensity $A(i, j)$	Time (s)
(a) DRR generation	Image/object ¹	Nonuniform	Penetrated voxels	993.7
(b) Gradient image generation	Image	Uniform	Surrounding $K \times K$ neighborhood intensities	67.2
(c) NCC computation	Image ²	Uniform	Corresponding intensity $B(i, j)$	2.5

1, 2: Communication is required to produce the final DRR and to perform reduction operations, respectively.

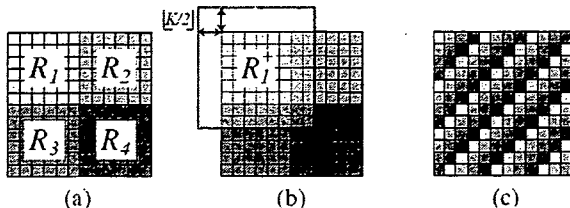


Figure 3. Workload distribution.

region, as shown in Figure 3. Here, the overlap size is given by the kernel size K , allowing processors to produce gradient images without any communication. As compared with other distribution schemes such as 1-D/2-D disjoint block and cyclic schemes, our scheme has the following advantage/disadvantages: (1) less communication, achieved by the overlap region; (2) more computation, due to the redundant DRR generation for the overlap region; and (3) imbalanced workload, as compared with the cyclic scheme.

If the overlap region is not given, communication is required for block boundaries in order to obtain intensities of neighbor points computed by other processors. This communication becomes a significant performance bottleneck in a case where many processors are responsible for the neighbor points. In this case, processors need to gather the intensities from many processors and also have to scatter their own intensities to others, but it is not easy to realize both fast scatter and gather operations at the same time. Due to this complex communication pattern, the cyclic scheme possibly results in poor performance. Furthermore, the 1-D/2-D block scheme without the overlap will also suffer from this situation as the number of processors P increases, because the kernel size K is relatively large compared to the block size, which decreases as P increases.

3.3. Proposed Method

We denote by $\mathcal{R} = \{(i, j) \mid 1 \leq i \leq S_I, 1 \leq j \leq S_J\}$ the domain of the ROI specified on the 2-D image, where S_I and S_J are the horizontal and vertical size of the ROI, respectively. Let \mathcal{R}_p , where $1 \leq p \leq P$, be the p -th subdomain partitioned by the 2-D disjoint block scheme such that $\mathcal{R} = \bigcup_{p=1}^P \mathcal{R}_p$ and $\mathcal{R}_p \cap \mathcal{R}_q = \emptyset$, for all $1 \leq p < q \leq P$. Let \mathcal{R}_p^+ be the p -th subdomain with its overlap region.

Given P processors, our parallel method aligns the volume V to the image I_F as follows.

1. Data load: For all $1 \leq p \leq P$, processor p loads the volume V from its local disk into main memory, and waits for registration tasks to be submitted. Then, processor 1 serves as a gateway receiving a registration task with its input data: the projected image I_F , the initial parameter \mathbf{T} , and the initial step size λ . After this, the gateway broadcasts the data to all processors.
2. DRR generation: For all $1 \leq p \leq P$, processor p locally generates a DRR for subdomain \mathcal{R}_p^+ .
3. Gradient image generation: For all $1 \leq p \leq P$, processor p locally generates the gradient images for disjoint subdomain \mathcal{R}_p .
4. NCC computation: For all $1 \leq p \leq P$, processor p locally computes six local sums from subdomain \mathcal{R}_p . Then, every processor participates in a reduction communication to combine local sums from all processors and distribute the global sum back to all processors. After this communication, every processor has six global sums, so that locally computes NCC.
5. Optimization: Repeat 2.–4. 13 times to update the parameter \mathbf{T} by using the steepest descent optimization. Repeat this step until a local optimum has been found.

4. Experimental Results

To evaluate the performance of our parallel method, we have implemented it using the C++ language and Message Passing Interface (MPI) standard.

We used a cluster of 64 PCs, each equipped with two Pentium III 1-GHz processors. The interconnection between nodes is a Myrinet switch, yielding a bandwidth of 2 Gb/s. Our implementation runs on a Linux operating system with the MPICH-SCORE library, a fast MPI implementation. **On timing results.** We performed registration tasks using datasets of a femur phantom and a real spine (see Table 2). The biplane images are generated as the front (coronal) view and the side (sagittal) view of the body. The kernel size K of the Gaussian filter was experimentally determined as $K = 19$ pixels ($\sigma = 3$).

We produced the CT volume and distributed it with its ROI information to each node before running our registration program. This distribution takes 1.7 and 3.8 s on the Myrinet network for the femur and the spine datasets, respectively. On the other hand, the 2-D fluoroscopy images are produced immediately before registration and then are broadcast by the registration program itself. It takes 37 and 119 ms to broadcast each dataset, respectively.

Table 2. Dataset specification.

	Femur phantom	Real spine
3-D volume size	256 × 256 × 367	512 × 512 × 204
File size	45 MB	102 MB
ROI size	53 × 47 × 54	299 × 299 × 47
2-D image size	640 × 512	1024 × 1024
File size	320 KB	2 MB
ROI size (front)	353 × 276	340 × 204
ROI size (side)	344 × 272	336 × 200

Table 3. Registration time and speedups.

# of procs <i>P</i>	Femur phantom		Real spine	
	Time (s)	Speedup	Time (s)	Speedup
1	320	1	1066	1
8	52	6	171	6
16	31	10	101	11
32	19	17	62	17
64	13	26	44	24
128	9	35	35	31

Table 3 shows experimental timing results on different numbers of processors. We can see that our implementation running on $P = 128$ reduces computation time for the spine dataset from 17 m (1066 s) to 35 s. It also achieves a shorter time of 9 s for the femur dataset with a smaller ROI. Times of less than 60 s are compatible with time constraints required for surgery. Thus, our parallel method enables us to utilize registration technique during surgery without degrading the quality of alignment.

On workload distribution. If we change our distribution scheme to a 1-D block scheme with overlap, the size of disjoint blocks and that of overlapping blocks on $P = 128$ become 3×276 and 21×294 pixels, respectively. Therefore, this 1-D scheme requires about 7.5 times more computation, resulting in a lower speedup. Moreover, since the vertical length of 1-D blocks becomes shorter than that K of the filter, processors need to communicate with more processors to obtain intensities of vertical neighbors, having a more complex communication pattern with network contention.

Although our overlapping scheme requires redundant computation for DRR generation, this disadvantage is covered by the advantage of less communication. If a 2-D disjoint block scheme is employed, every processor needs to communicate its boundary data with its vertical/horizontal/diagonal neighbors. Though this can be implemented by repeating shift communication operations, these operations could be a performance bottleneck. For example, when using 128 processors for the femur dataset, this scheme causes 2.7 KB ($41 \cdot 53 - 23 \cdot 35$ pixels, each in 2 bytes) of incoming data and the same amount of outgoing data at every processor, which must be sequentially processed in eight shift communication operations.

On speculative processing. In our experiments, we found that the speedup was limited by a relatively smaller value,

as compared with P . In this situation, where the speedup is theoretically limited by a small value, using more processors for data-parallel processing results in a lower utilization of computing nodes. To deal with this, our method tries to raise the speedup by means of speculative processing. This strategy will lead to a higher speedup if registration tasks are repeatedly processed with different initial parameters due to unsuccessful alignments.

Another important motivation to exploit speculative parallelism comes from the fact the registration algorithm sometimes fails to align objects due to local optimums. Therefore, our strategy will also improve the confidence of registration technique, providing more robust alignment for a wide variety of clinical scenarios.

5. Conclusions and Future Work

We have presented a parallel method for 2-D/3-D registration, aiming at realizing intraoperative alignment. Our method exploits data and speculative parallelism in an intensity-based algorithm, so that can perform fast, accurate, and robust registration during surgery. Our implementation on a cluster of 64 PCs aligns a $299 \times 299 \times 47$ voxel volume to 340×204 pixel images in a few tens of seconds, a clinically compatible time.

In the future, our parallel implementation could be improved by exploiting task parallelism in order to achieve further acceleration.

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Communications

Fluoroscopic Bone Fragment Tracking for Surgical Navigation in Femur Fracture Reduction by Incorporating Optical Tracking of Hip Joint Rotation Center

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Abstract—A new method for fluoroscopic tracking of a proximal bone fragment in femoral fracture reduction is presented. The proposed method combines 2-D and 3-D image registration from single-view fluoroscopy with tracking of the head center position of the proximal femoral fragment to improve the accuracy of fluoroscopic registration without the need for repeated manual adjustment of the C-arm as required in stereo-view registrations. Kinematic knowledge of the hip joint, which has a positional correspondence with the femoral head center and the pelvis acetabular center, allows the position of the femoral fragment to be determined from pelvis tracking. The stability of the proposed method with respect to fluoroscopic image noise and the desired continuity of the fracture reduction operation is demonstrated, and the accuracy of tracking is shown to be superior to that achievable by single-view image registration, particularly in depth translation.

Index Terms—Fluoroscopic bone fragment tracking, fracture reduction, kinematic knowledge of the hip joint, motion constraint.

I. INTRODUCTION

Computer guidance for femur fracture reduction has been studied as a means of improving repositioning accuracy and reducing radiation exposure to medical staff by shortening procedures and introducing robotic surgery. Where previous studies have focused on femoral shaft fractures, proximal femur fractures such as femoral intertrochanteric fractures are also common, particularly in osteoporotic elder females. In previously proposed systems, reference markers for the positional

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sensor are attached to both proximal and distal bone fragments to measure the spatial relationship between the fractured bone fragments. In a proximal femur fracture, however, it is difficult to attach the reference marker to the proximal fragment, as the fragment is located deep in the body and the implantation required for attachment may have an undesirable effect on the hip joint in leg position changes during surgery.

Some researchers have investigated a fluoroscopic tracking method for bones [1], [2]. The methods involve the use of 2-D/3-D registration, which in conventional single-view procedures does not provide sufficient accuracy in depth translation for clinical use of the fracture fragment tracking. Although stereo-view 2-D/3-D registration overcomes some of the shortcomings of single-view registration, interactive C-arm positioning is required in each registration step. The method proposed by the present authors effectively combines 2-D/3-D registration from a single-view fluoroscopic image with the proximal femoral fragment position obtained by pelvis tracking. The use of such a 2-D/3-D registration technique by incorporating positional information at a number of points was originally proposed by Russakoff [3] as a means to improve the accuracy of stereo-view registration. However, in that pioneering study, fiducial markers were embedded directly in the bone to acquire the positional information. In contrast, our method employs pelvis position tracking as a less-invasive approach, and knowledge of the kinematics of the hip joint is applied to acquire the position of the femoral head center. The proposed method therefore does not require the implantation of a fiducial marker in bones such as the proximal femur fragment.

II. METHOD

A. Process Overview

The segmentation of the pelvis and fragments of the fractured femur are processed preoperatively using a computed tomography (CT) volume and in-house segmentation software. The femoral head center, employed as the hip joint rotation center, is localized by sphere-fitting image processing in the CT volume coordinate system. Bone tracking during surgery is performed in two registration steps. In the first registration step, two imaging operations and a position measurement procedure are performed once for conventional stereo-view 2-D/3-D registration to determine the initial pose of the segmented bone volume and the geometric relationship between the pelvis reference marker and the hip rotation center point. Single-view fluoroscopy imaging and a single position-tracking procedure are then performed for each iteration of the hybrid bone tracking described below. This procedure is performed iteratively for bone tracking during surgery.

B. Hybrid Bone Tracking Method

The method employs two constraints: an image similarity measure, and a positional constraint for the hip joint rotation center based on optical tracking. Gradient correlation is employed as an image similarity measure, and is sensitive to thin-line intensity differences such as bone edges, although it is potentially insensitive to the overlap of soft tissue.

Let E_{image} represent the evaluation value for image similarity measure. The positional constraint of the hip joint rotation center can then be expressed as

$$E_{\text{position}} = \text{dist}(p', p_{\text{ref}}). \quad (1)$$